Proactive and reactive adaptability of elderly adults with respect to dynamic stability

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Stefanie Bierbaum

Präsident der Humboldt-Universität zu Berlin: Prof. Dr. Jan-Hendrik Olbertz

Dekan der Philosophischen Fakultät IV: Prof. Dr. Dr. hc. Jürgen van Buer

Gutachter:

1. Prof. Dr. Adamantios Arampatzis
2. Prof. Dr. Urs Granacher
3. Assoc. Prof. Dr. Uwe Kersting

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Zusammenfassung

Hindernisse oder Untergrundveränderungen während des Gehens sind eine große Herausforderung für das menschliche System. Die entsprechenden Anforderungen sind für ältere Personen im Vergleich zu jüngeren jedoch durch die altersbedingte Beeinträchtigung afferenter und efferenter Systeme höher. Das zeigt sich unter anderem in dem erhöhten Sturzrisiko älterer Personen in Folge von unerwarteten Störungen. Das menschliche System ist allerdings in der Lage, sensomotorische Adaptationen entsprechend aktueller Anforderungen und früherer Erfahrungen durchzuführen. Die vorliegende Arbeit versucht daher, Erkenntnisse bezüglich der prädiktiven und reaktiven Anpassungsfähigkeit älterer Personen während der Fortbewegung zu erlangen. Die Anpassungsfähigkeit auf unerwartete Gangstörungen wurde mittels eines Gangstegs untersucht, in welchen ein austauschbares, für die Probanden nicht sichtbares, Element eingelassen war. Der Gang jüngerer und älterer Probanden zeigte ähnliche prädiktive Anpassungen. Im Gangversuch direkt nach der ersten unerwarteten Störung erhöhten beide Altersgruppen ihre dynamische Stabilität zum Zeitpunkt kurz vor der erwarteten Gangstörung im Vergleich zu unbeeinflusstem, normalem Gang, was auf schnelle prädiktive Anpassungen hinweist. Dieses Ergebnis bedeutet, dass Personen ihre Fähigkeit zur prädiktiven Anpassung der dynamischen Stabilität als Reaktion auf Gangmanipulationen auch im Alter beibehalten können. Durch die wiederholte Erfahrung unerwarteter Störungen zeigten beide Altersgruppen eine Verbesserung ihres Stabilitätszustandes infolge der Gangstörung. Nach der fünften Gangstörung war eine deutliche reaktive Adaptation der Stabilität an die Störung zu sehen. Ältere Personen zeigten allerdings eine Tendenz zu einer geringeren Adaptation verglichen mit den jüngeren. In einer Interventionsstudie wurde des Weiteren untersucht, ob ältere Personen generell verbesserte Reaktionen auf Stabilitätsstörungen lernen können. Zwei Trainingsgruppen nahmen für 14 Wochen, zweimal pro Woche (á 1.5h) an einem Trainingsprogramm teil und führten Übungen durch, die Mechanismen der dynamischen Stabilität enthielten. Die Stabilitätstrainingsgruppe (ST) trainierte diese Übungen in der gesamten Übungszeit, während die Trainingseinheiten der kombiniert trainierenden Gruppe (MT) zusätzlich Krafttrainingsübungen für die unteren Extremitäten enthielten. Die Kontrollgruppe führte kein Training durch und zeigte auch keinerlei Veränderungen in der Stabilität im Vergleich prä- zu post-Messung. Nach der Intervention war für ST die dynamische Stabilität nach der unerwarteten Gangstörung signifikant erhöht im Vergleich zum Stabilitätszustand vor der Intervention. Beide Interventionsgruppen vergrößerten ihre Unterstützungsfläche infolge der Gangstörung nach der Intervention verglichen mit der Größe der Unterstützungsfläche vor der Intervention. Signifikante Unterschiede waren hier allerdings nur bei ST zu finden. Aus diesen Ergebnissen kann geschlossen werden, dass das Training von Übungen, die die Mechanismen der dynamischen Stabilität enthalten, zu einer besseren Anwendung dieser Mechanismen nach einer unerwarteten Gangstörung führt. Das kombinierte Trainingsprogramm zeigte allerdings keine Vorteile gegenüber dem ausschließlichen Training der Mechanismen der dynamischen Stabilität.

Abstract

Coping with obstacles or unexpected surface changes during walking is a big challenge for the human system. The demands, however, are even higher for older compared to young subjects because of the age-related deterioration of the afferent and efferent systems. This is evident in a higher fall risk for the elderly in consequence to unexpected perturbations. Yet, the human system is able to show sensorimotor adaptations according to actual demands and in consequence of prior experience. This thesis therefore aimed to gain knowledge about the preservation of predictive as well as reactive adaptability in old adults during locomotion. Adaptability to unexpected perturbations during walking was investigated by the application of a walkway, which included an exchangeable invisible element.

Gait of young and old adults revealed similar predictive adjustments. Both age groups increased their dynamic stability at touchdown of the disturbed leg (prior to the perturbation) in the trial following the first unexpected perturbation compared to unaffected normal gait, reflecting fast predictive adjustments. This suggests that older adults preserve the ability to show predictive adaptations of their dynamic stability in consequence to perturbations during walking.

In the course of several unexpected perturbations, both age groups showed an increase in their dynamic stability state after the perturbation. Considerable reactive adaptations of the dynamic stability to the perturbation were seen in consequence of the fifth perturbation. Older adults, however, showed a tendency towards a lower adaptation magnitude compared to young adults.

Furthermore, by means of an intervention study, this thesis investigated if older adults preserve their ability to learn improved postural reactions in consequence to perturbations in general. Two training groups participated for 14 weeks, twice a week (á 1.5h), in a training program and performed exercises which included mechanisms of dynamic stability. The stability training group (ST) exercised those tasks for the entire training volume whereas the mixed training group (MT) additionally performed strength training exercises for the lower extremities during the training sessions. The control group performed no training program and accordingly showed no differences in the stability between pre and post measurements. Post-intervention, dynamic stability state was significantly increased for the ST group compared to pre-intervention after the unexpected perturbation during walking. Both intervention groups increased their base of support of the recovery step in consequence to the perturbation after the intervention, showing significant differences only in the ST group. Therefore, it can be concluded that exercising the mechanisms of dynamic stability led to a better application of these mechanisms after an unexpected perturbation during gait. The mixed training program, however, shows no advantages compared to the exclusive training of the mechanisms of dynamic stability.

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1. Introduction

Human biped locomotion even in young and healthy individuals can be seen as a movement of an instable and fragile system. Locomotion has to fulfill several complex tasks such as the generation of progression, maintenance of equilibrium, adaptability to changes and the initiation and termination of locomotor movements *(Woollacott & Tang 1997; Shumway-Cook & Woollacott 2007)*. These tasks have to be fulfilled while handling with the characteristics of the human bipedal gait. The human system not only has to handle with a small and from time to time changing base of support, but also with the location of two-thirds of the body mass in a height of two-thirds of the body height *(Winter et al. 1990)* which implies a high potential energy. Therefore, the behavior of the whole body system during gait may be compared with an inverted pendulum. The requirements concerning stability are high even during constant gait in a healthy body, but are more demanding in anticipation of distinct external conditions, after perturbations or as a consequence of impaired internal systems. Especially the group of elderly adults shows problems with postural stability and reveals an increased fall risk.

The ability to control balance declines with increasing age due to the deterioration of sensory, motor and cognitive systems. Impairments in one of those systems or in a combination of them may result in a less coordinated handling of everyday challenges. In addition to the age-related deterioration of those systems, medication and psychological factors like fear of falling contribute to the actual and experienced stability *(e.g. Deandra et al. 2010; Delbaere et al. 2010)*. Regarding the fall risk of elderly it is important to identify the affected subsystems and to remedy any functional impairment if possible. In addition, it is important to identify the situations in which elderly adults fall. The knowledge about potential precarious situations may help to either avoid the exposure to these hazards or alternatively to learn how to handle those challenges. During daily life, behavior often has to be modified because of new, unknown or changed situations or because of altered intrinsic properties of the individual. Avoidance of the exposure to altered conditions seems to be impossible and therefore it is essential to preserve adaptability with increasing age. A preserved adaptability may help to handle with environmental hazards in advance. In anticipation of environmental hazards the motor behavior may be adapted at its best to the actual requirements. However, not all environmental hazards can be identified beforehand – some situations demand the ability to adapt rapidly and adequately in response to experienced perturbations. Especially the reactive behavior in consequence to perturbations has shown to be impaired in the elderly *(Thelen et al. 1997, 2000; Grabiner et al. 2005; Pijnappels et al. 2005; Karamanidis & Arampatzis 2007)*. Therefore, it is important to know if elderly adults are able to learn better reactive responses in consequence to perturbations and if this reactive adaptability is not only specific to particular perturbations but shows also generalization of the ability to cope with unexpected perturbations. If this reactive adaptability shows to be universally applicable, training regimens could practice the reactive behavior in order to achieve a general improvement afterwards perturbations.

This thesis tries to clarify if elderly adults show a preserved adaptability with respect to their dynamic stability. This thesis consists of three studies, all investigating the sensorimotor adaptability of elderly. The first study is designed to investigate the predictive, short-term adaptability of elderly compared to young adults in response to several perturbations during walking. The second study investigates the reactive, short-term adaptability of elderly in comparison to young adults to unpredictable perturbations during locomotion. Finally, the third study examines the possibility of long-term improvement in the application of postural strategies due to training of specific underlying mechanisms of dynamic stability.

2. Literature Review

2.1 Fall risk in the elderly

Falling and its resulting injuries are a growing public health concern with estimations of one third of community-dwelling adults over 65 years of age which will experience one or more falls every year *(Blake et al. 1988; da Cruz et al. 2012)*. However, the problem in the elderly population is not only the high incidence of falls, but the combination of falls and a high susceptibility to injury *(Rubenstein 2006)*. This suggests that fall related injuries are an important problem in this age group *(Blake et al. 1988; Tinetti et al. 1988)*. According to *Heinrich et al. (2012)*, falls contribute to a high amount to the overall cost of injuries in elderly. Hip fractures of German nursing home residents cause estimated costs of about 8160 Euro each and the overall costs of falls in elderly account for about 2.1 – 3.4 billion Euro per year *(Heinrich et al. 2012)*. Of course, not only the expanding costs for the health system are a challenging problem, but also the social and individual problems which are accompanied by falls.

In consequence of the development of birth- and death rates, there will be a significant change in the demographic structure in Germany and other states. In Germany, the percentage of people in the age above 65 years is estimated to grow from 16.7 % and 20 % for men and women to 22.3 % and 29 % respectively. This means, that the age group of individuals in the age of 65 years or older will grow for about one third (33%) from 16.7 million in 2008 to 22.3 million persons in the year 2030 *(Statistisches Bundesamt 2011)* and thereby the problem of falls will grow accordingly.

Therefore, the diminishment of falls and fall related injuries is a challenging issue in our society. It is essential to identify age-related changes in the physiological, neuromuscular, sensory and cognitive systems to assess the ongoing aging process and further to assess postural instability in an early stage. In addition, sustained resources and abilities in the age have to be identified. The identification of predictors for successful motor and cognitive aging will enable us to create successful intervention strategies to counteract age related impairments.

One model for the explanation of decreasing performance in postural control assumes that the age-related degeneration of the musculoskeletal, neuromuscular and sensory systems causes the postural instability in older adults *(Horak et al. 1989a; Woollacott 1989)*. Another model for the increased instability with age, though, assumes a quite small effect of age per se on postural control and postulates an influence of pathologies which lead to the degeneration of different systems *(Horak et al. 1989a; Woollacott 1989)*. The reason for the age-related degeneration in physiologic systems may be not clear until now, but the implication of degenerated systems may be an altered postural stability.

Another model for the process of withdrawal from hazards to control balance is shown in *figure 2-1*. This scheme, adopted from *Wijlhuizen (2008)*, illustrates the age-related change of the level of balance control capability and the corresponding variability. With ageing, the initial high capability to control balance and the associated low variability change: the capability to control balance diminishes and the variability increases, functioning less reliable. In addition, the level of demand of balance control decreases across the life-span together with its variability. The demands of balance control relate to the everyday challenges. During the early phase, balance control demands are quite high and highly variable, but show a gradual reduction due to the self-perception of a decreased capability to control balance. The lower confidence in the ability to cope with balance demands is compensated by a correspondent avoidance of several, more demanding, activities. Therefore, with growing avoidance of specific activities, the variability in the balance control demands reduces. With growing age, the safety margin between the level of balance control capability and the level of balance control demand diminishes, i.e. the probability of falls increases. This model helps to understand the relationship between the daily demands on balance control and the balance control capability *(Wijlhuizen 2008)*.

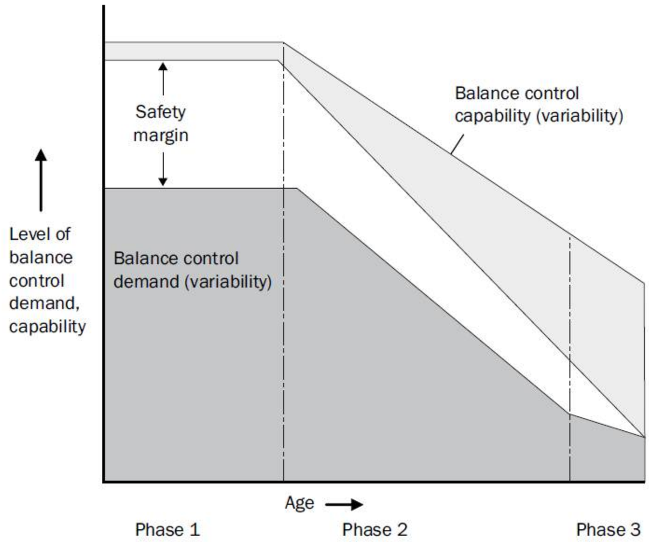

Figure 2-1: *Schematic representation of hypothesized level of balance control capability of a person and its variability (dark grey area), the level of balance control demand and its variability (light grey area), and the difference between them (the safety margin) for three age phases. Reprinted from Wijlhuizen (2008, p.124) with kind permission.*

Falls, in general, are a consequence of a complex interaction between extrinsic factors related to the personal environment and surroundings, task-related factors and intrinsic factors which are specific to each individual. Task-related factors refer to properties of the specific tasks such as complexity or speed. Intrinsic fall risk factors are, for instance, impairments in the sensory systems, reduced muscle strength, medication or diseases and further age and gender. With increasing age, intrinsic risk factors become more and more important *(Tideiksaar 2000)*. Therefore, the contribution of several components of the human body and the potential modification of their function with ageing has to be considered.

However, stability is not only a consequence of the functioning of single components of the (loco-)motor system, but moreover dependent on the interaction and coordination of different components and also significantly biased by the chosen postural strategy. For the control of balance, the sensory (afferent) as well as the neuromuscular (efferent) system both have considerable impact on postural control *(fig. 2-2);* both are concerned by ageing. During gait, specific patterns of muscle activity are modulated according to sensory information, external conditions and biomechanical constraints

(Tseng et al. 2009). Thus, impairments in the sensory and neuromuscular system not only influence the walking pattern but also influence the handling of unknown or unexpected situations and places older individuals at a higher risk for slip and fall accidents *(Lockhart et al. 2005)*.

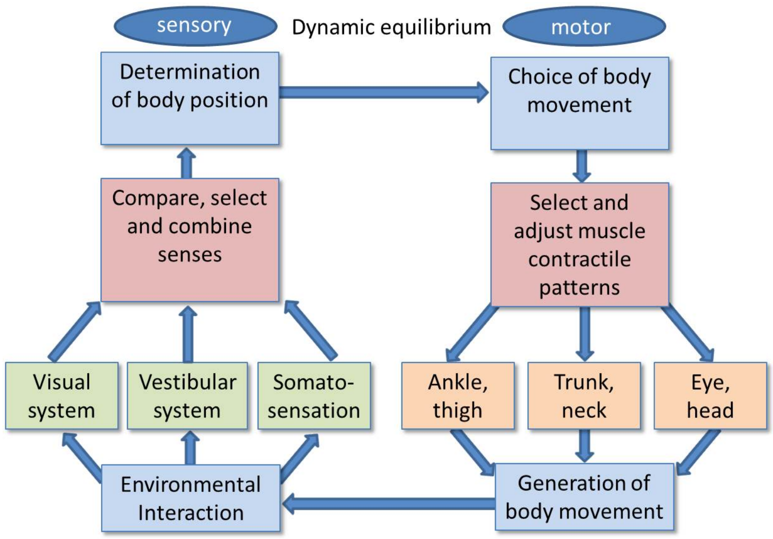

Figure 2-2: *Afferent and efferent subsystems which contribute to balance control. The sensory systems provide information about the body position and the environment. Body movement takes place when muscle contractile patterns are chosen and executed. According to the dynamic equilibrium model of Nashner (1990) in Spirduso et al. (2005, p.136) with kind permission of Springer Science + Business Media and Human Kinetics.*

2.2 Degeneration with ageing

2.2.1 Impairment of sensory systems

The sensory system related to postural control is relevant for the acquisition of adequate information about the position of the body and the center of mass in relation to the support surface. Sensory systems which affect balance are the somatosensory, the visual and the vestibular system *(Spirduso et al. 2005; Peterka 2007; Shumway-Cook & Woollacott 2007)*.

The *somatosensory system* provides sensory information about the motion of the body with respect to the support surface and the motion of limbs with respect to each other *(Horak et al. 1989; Sturnieks et al. 2008).* It includes receptors in the joints, tendons and muscles. These receptors provide feedback about joint position, muscle length and tension, movement velocity and touch. Aging seems to be accompanied by reduced static and dynamic muscle spindle sensitivity *(Miwa et al. 1995)* as well as by decreased vibration perception *(Verrillo et al. 2002)* and touch threshold *(Perry 2006)*. These changes are caused by a decreased number of receptors *(Bolton et al. 1966; Swash & Fox 1972)*, a desensitization of muscle spindles *(Mynark & Koceja 2001)* and a demyelination of sensory axons *(Verdú et al. 2000)*. Furthermore, age may lead to a diminished joint position sense *(Petrella et al. 1997; Hurley et al. 1998; Goble et al. 2009; Ribeiro & Oliveira 2009)*. This impaired function of the somatosensory system provides reduced information about the position of the limbs and therefore decreases movement precision and postural control *(Hay et al. 1996; McChesney & Woollacott 2000)*. The deterioration of the proprioceptive receptors has been associated with increased postural sway *(Lord et al. 1991; McChesney & Woollacott 2000)*, impairments in the performance on functional tasks *(Hurley et al. 1998)* and eventually with an increased fall risk of old adults *(Woollacott et al. 1986; Horak et al. 1989).*

The *visual system* delivers continually updated information regarding the position and movement of body segments relative to each other and the extrapersonal space. With increasing age, the visual system undergoes physiological changes and shows a decline in several visual processes such as visual acuity, contrast and glare sensitivity, dark adaptation, accommodation and depth perception *(for review see Sturnieks et al. 2008)*. The reduced performance of those functions has shown to be associated with an increased fall risk *(Ivers et al. 1998, 2000; Lord & Dayhew 2001)*. This may be explained by a misjudgment of distances and misinterpretation of spatial information through impaired visual input. Despite the deterioration of those functions with aging, old adults show an increased dependence on visual inputs for the control of balance *(Horak et al. 1989).* Therefore, visual cues become more and more important for the postural control of elderly *(Sundermier et al. 1996).*

Aging of the *vestibular system,* which generates perception about the linear and angular acceleration of the head, is accompanied by attrition of neural and sensory cells in the peripheral labyrinths *(Herdman et al. 2000).* The vestibular system provides an orientation reference for the other sensory systems *(Keshner & Cohen 1989)* – therefore, the age-related alterations in the vestibular function may decrease the reliability of this reference and may cause problems when integrating information from the other sensory systems *(Teasdale et al. 1991)*. The vestibular contribution to postural control depends on the type of perturbation which is imposed as well as on the importance of this information in the integration of all sensory systems. Individuals with manifested vestibular disorders may show no increased fall incidence due to compensatory effects of the other sensory systems and adopted corrective strategies *(Whitney et al. 2000; Baloh et al. 2001)*. Acute vestibular disorders, however, which lead to dizziness, may increase the fall risk.

2.2.2 Impairment of efferent systems

As mentioned above, not only the components of the afferent or sensory system are important influencing factors with regard to static and dynamic stability, but also the diverse components of the neuromuscular system. With ageing, several subsystems develop impairments in their functionality.

For the muscular system, muscle mass and contractile qualities are capacity determining factors. The observed loss in muscle mass with ageing *(Lexell 1995; Narici et al. 2003; Hunter et al. 2004; Sturnieks et al. 2008)* can partly be explained by hormonal, immunologic and myocellular causes as well as by decreased muscular activity and a reduced protein intake with age *(Vandervoort 2002; Manini & Clark 2012)*. The muscle atrophy can be observed in a loss of muscle fibers and a reduction in the muscle fiber size, mostly of type 2 fibers *(Lexell et al. 1988)*. Age-related changes in the muscle fiber type composition *(Lexell 1995; Hunter et al. 2004)* and in the muscular architecture *(Narici et al. 2003)*, which includes changes in fascicle length and pennation angle as well as age-related changes in the excitation-contraction coupling processes *(Delbono et al. 1995)* effect the decreased contractile quality in age *(Thelen et al. 1996; Manini & Clark 2012)*. As a consequence of these changes and an additional decrease in tendon stiffness *(Karamanidis & Arampatzis 2005; Onambele et al. 2006)*, the force-velocity relationship of human muscles is modified with age and shows a decline in the force producing capacity and contraction velocity *(fig. 2-3; Hortobagyi et al. 1995; Thelen et al. 1996; Raj et al. 2010)*.

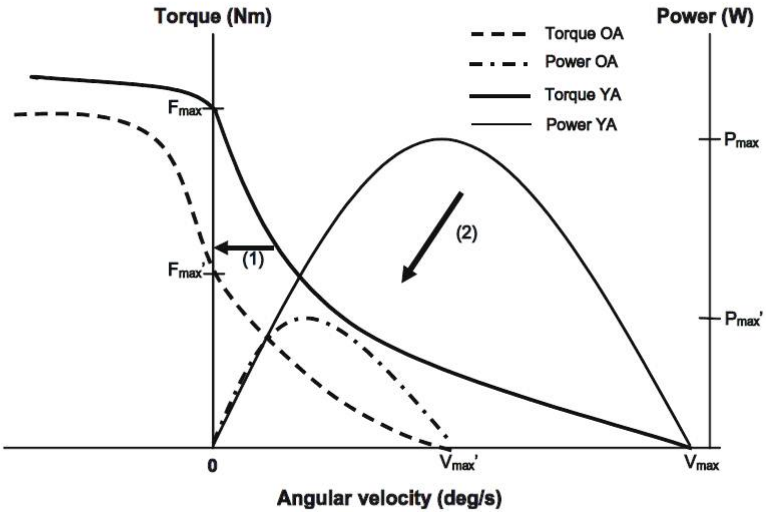

Figure 2-3: *Changes to the force-velocity- and power-velocity relationship with age. With ageing, maximum contractile velocity as well as the maximum voluntary force is decreased. OA: old adults, YA: young adults. Reprinted from Raj et al (2010, p.85) with permission by Elsevier.*

With ageing, the size of the motor unit decreases as well as the number of excitable motor units and the maximal motor unit discharge frequency *(Brown et al. 1988; Doherty et al. 1993; Vandervoort 2002; Klass et al. 2008)*. Reduced supraspinal drive and decreased spinal excitability with the aging process *(Manini & Clark 2012)* may further contribute to a diminished muscle performance. Those physiological and structural changes altogether affect the muscular strength and power across all contraction speeds *(Larsson et al. 1979; Wolfson et al. 1985; Thelen et al. 1996; Trappe et al. 2003; Thom et al. 2005, 2007; Pijnappels et al. 2006; Raj et al. 2010).* The reduced muscle torque and power in the elderly can also be explained by the increased coactivation of the antagonist muscles which causes limited movement efficiency *(Izquierdo et al. 1999; Macaluso et al. 2002; Benjuya et al. 2004).* On the other hand, the increased coactivation in elderly may protect and stabilize the joint during forceful contractions *(Macaluso et al. 2002)*. The enhanced coactivation in elderly is suggested to be the consequence of an impaired regulation of inhibitory Ia-Interneurons through supraspinal centers *(Morita et al. 1995; Mynark & Kocera 2002)*.

The combined loss of muscle mass and muscle strength, accompanied by low physical performance is defined as "*sarcopenia" or "dynapenia"*. In contrast to the original definition of sarcopenia, the actual working definition of sarcopenia includes not only loss of muscle mass, but also loss of muscle strength *(Cruz-Jentoft et al. 2010; Manini & Clark 2012)*. Sarcopenia, which is often said to be one of the most important causes for falls *(Landi et al. 2012),* may originate from some of the aforementioned factors. Decreased muscle strength, especially in the lower extremities, has shown to an independent risk factor for falls *(Moreland et al. 2004)*.

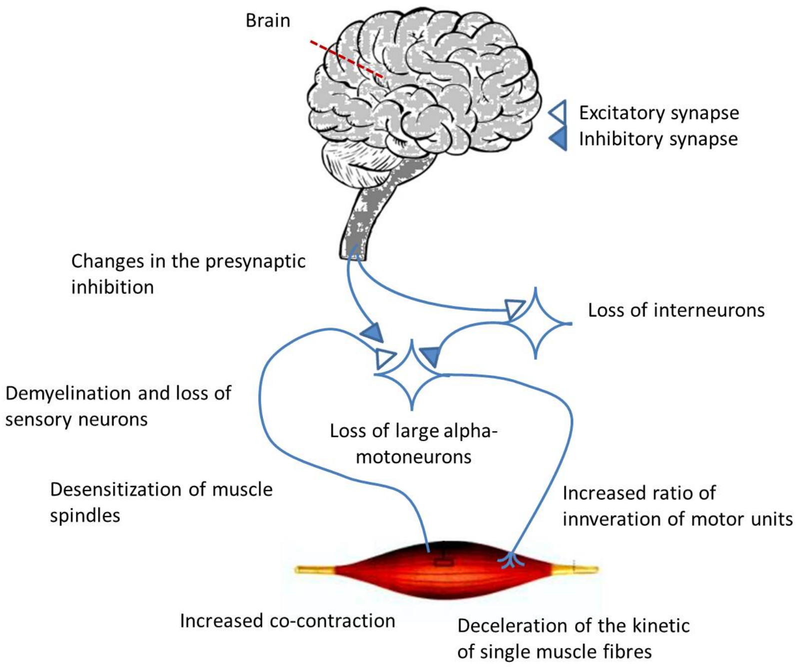

Figure 2-4: *Age-related changes in the sensorimotor system. Changes in different places of the sensorimotor system (excitatory as well as inhibitory systems) contribute to age-related impairments like desensitization or the slowing down of neuromuscular performance. Adapted from Granacher & Gollhofer (2005, p.70) with permission by the Deutsche Zeitschrift für Sportmedizin.*

Eventually, the *interplay* of the afferent and efferent system may be impaired by modifications in the different components of the postural control system, which contributes to an impaired reflex and recovery behavior in the elderly. The response latency of elderly, for example, is increased by changes in the receptors (desensitization of muscle spindles) and in the afferent neurons which conduct the signals to higher centers (e.g. demyelination) *(Swash & Fox 1972; Mynark & Koceja 2001)*. Reduction in the explosive force production may be a consequence of a decreased number of large alpha-motoneurons and interneurons *(Terao et al. 1996)* and may also be the result of an altered recruitment and discharge frequency behavior of motor units *(Kamen et al. 1995)*. The interplay and the point of application of those factors in the sensorimotor system can be seen in *figure 2-4*.

2.2.3 Impairment of cognitive systems

An essential factor for the generation of adequate reactions in response to perturbations is the ability to generate the postulated behavior in an appropriate time frame. With ageing, individuals show an increased reaction time at simple reaction tasks, even more pronounced at complex reaction tasks *(Fozard et al. 1994; (Der & Deary 2006; Eckner et al. 2012)*. This may be the consequence of a diminished processing of information and has been observed in deficits in stimulus encoding, information integration, central processing as well as response initiation and preparation *(Salthouse & Somberg 1982; Stelmach & Worringham 1985; Young & Hollands 2012)*. Increased reaction time with age has been shown to be associated with an increased risk of falls *(Lord et al. 1992; Lajoie et al. 2002; Lajoie & Gallagher 2004)*. However, most tests for the assessment of reaction time are based on cognitive reaction time tasks and it is not clear if the observed increase in reaction time is of significance in most tasks of daily living. Measuring the (motor) reaction time to dynamic, balance recovery situations, for instance, revealed no differences between young and old adults and fallers and non-fallers *(Arampatzis et al. 2008, 2011)*. In general, older adults have shown to be slower in initiating voluntary actions compared to young, but are as quick as young when stepping reactions are evoked by postural perturbations *(Luchies et al. 1999; Rogers et al. 2003)*. These changes in the information processing speed of elderly may be attributed to changes in the central and peripheral nervous systems. For example, there are indications for a less coordinated activity in the brain of elderly, suggesting a global loss of integrative function *(Andrews-Hanna et al. 2007)* and a less localized neural activity in some brain regions involved in executive functions *(Park & Reuter-Lorenz 2009)*. This delocalization, a combination of over- and underactivation in elderly, can be explained either by the de-differentiation or by the compensation view. The de-differentiation view suggests that old adults inefficiently recruit additional brain regions because of a less precise brain structure-function relationship *(Riecker et al. 2006)*. The compensation view assumes that the additional recruited brain regions compensate for structural and biochemical declines *(Mattay et al. 2002)*. Recent theories propose a scaffolding theory of aging and cognition whereas the adaptive brain reorganizes according to challenges which are posed by declining neural structures and functions *(Park & Reuter-Lorenz 2009)*.

Structurally, the cerebral cortex shows a reduction of grey and white matter in the prefrontal cortex and a mass reduction of the frontal lobe with ageing *(Raz et al. 1997, 2005; Gunning-Dixon et al. 2009; Juraska & Lowry 2012)*. Age is accompanied by a loss of neurons, dendrites and synapses and changes in neurotransmitter systems and there are indications for a loss of myelin in the central nervous system *(Juraska & Lowry 2012)*, which may be interrelated with the slowing down of the central processing. Executive functions and working memory performance, both believed to be located in the frontal lobes, are most vulnerable to the age-related decay *(Salthouse 1994; West 1996; Gunning-Dixon & Raz 2003)*. Loss of executive function is observable in a decreased planning or self-monitoring ability, problems in modifying the behavior in consequence to changing task demands, attentional dysregulation and perseveration to conservative patterns *(Gunning-Dixon & Raz 2003)*. The decline in executive control function is thought to stem from the inefficiency of inhibitory processes *(Park & Reuter-Lorenz 2009)*. *Figure 2-5* shows the age-related gradual declines in the cognitive mechanisms of speed, working memory and long-term memory *(Park & Reuter-Lorenz 2009)*.

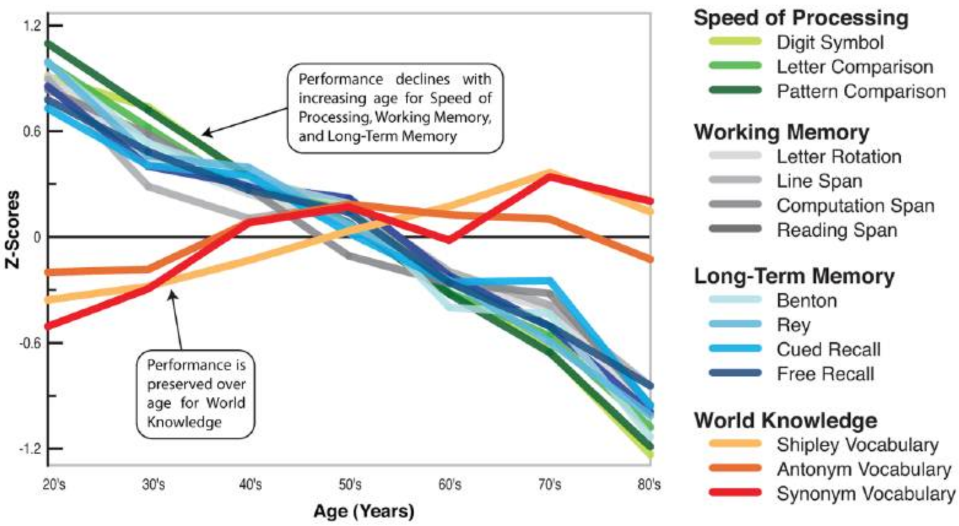

Figure 2-5: *Cross-sectional aging data showing performance on speed of processing, working memory, long-term memory and world knowledge*. *Reprinted from Park & Reuter-Lorenz, (2009, fig. 1, p.196) with permission by Annual Reviews.*

Cognitive impairment and dementia have been associated with an increased fall rate *(Shaw 2002)*. This supports recent studies which suggest a link between muscle weakness and cognitive decline *(Auyeung et al. 2008; Boyle et al. 2009)*. It is suggested that loss of muscle strength could also be the result of an underlying disease process that also leads to cognitive decline *(Boyle et al. 2009)*. Furthermore, increasing evidence links executive function and attention with movement, suggesting that cognitive impairment may contribute to motor decline *(Yogev-Seligman et al. 2008; Boyle et al. 2009)*. Executive function seems to contribute to the regulation of gait in older adults, especially during complex and challenging conditions *(Mirelman et al. 2012)*. Therefore, deficits in executive function may increase the risk of falls. The performance in attention and executive function tests, for instance, allowed the prediction of future falls *(Mirelman et al. 2012)*.

The aforementioned changes in the several systems with ageing contribute, depending on the observed system, more or less to balance disorders. In general, sensory subsystems generate redundant information – and therefore, impairment of one system can be compensated by the others. However, the ability to weigh conflicting information from different sensory subsystems may also be modified with ageing *(Shumway-Cook &*

Woollacott 2000). For a significant analysis of the impact of the various factors it is essential to investigate the parameters which are important for dynamic stability.

2.3 Dynamic stability in the elderly

2.3.1 Dynamic stability control

2.3.1.1 Predictive and reactive motor behavior

Postural control is essential for all voluntary movements and naturally also for the control of more or less automatic postural responses. Postural control therefore is a requirement for all movements and tasks – however, the stability and orientation demands for postural control change according to the task and the intention of the subject *(Shumway-Cook & Woollacott 2007)*. In contrast to earlier views, where balance was thought as a reflex-determined system in which only reactive responses to stimuli were possible, recent investigations assume a more complex and modifiable control of balance. Postural control has shown to be able to act *proactive* and *adaptive* and therefore there are indications for a, at least partially, central organization on the basis of prior experience and intention *(Horak et al. 1997)*.

There are two mechanisms for the control of stability: *predictive* or proactive mechanisms and *reactive* mechanisms which are feedback-based. The predictive mechanisms are based on a feed-forward movement plan and rely on knowledge about the environment which was generated by earlier experiences or which is available prior to the execution of the intended movement. Those mechanisms are utilized in predictable situations. The predictive control of movements is thought to be regulated mainly by the cerebellum *(Ramnani 2006; Bastian 2006)*. It is assumed that the cerebellum is provided with a copy of the motor commands, the so-called efference copy, which is used as an input to a forward model. The forward model uses the efference copies to predict the new state of the body after executing the motor commands and further to predict the sensory consequences *(Ramnani 2006)*.

Additional predictive behavior is suggested to be inherent to muscle spindles – type Ia muscle spindle afferents have shown to predict the future kinematic state of their parents muscle during active motor behavior *(Dimitrou & Edin 2010)*. According to *Woollacott and Tang (1997)*, proactive control mechanisms function in two ways: in the first way the proactive control is integrated for example into the normal walking pattern – i.e. in the form of muscle activation to reduce the inherent biomechanical threats to stability during walking. The second way of proactive control refers to an early detection of environmental hazards and the implementation of postural and locomotion adjustments prior to the contact with the hazard *(Woollacott & Tang 1997)*.

Reactive mechanisms are generated by the use of sensorimotor feedback and therefore depend on knowledge which is received during the movement. Those reactive mechanisms may be automatic (reflexes) or volitional (e.g. stepping response) and aim at the modification of movements which are already in progress. This modification may be necessary because of an incorrect feed-forward plan or the development of unpredictable perturbations *(Tseng et al. 2009)*. As mentioned before, the cortical contribution to reactive responses is still discussed controversial – in general, subcortical neural centers such as spinal cord or brain stem are thought to be responsible for the reactive adjustments *(Morton & Bastian 2006).* Yet, even the reactions to new, unexpected perturbations are not purely reactive, since expectation, attention, intention and the environmental context, together with preprogrammed muscle activation patterns (synergies), influence the reactive responses *(Horak et al. 1997)*.

2.3.1.2 Mechanisms and strategies for the control of dynamic stability

Regarding the preprogrammed muscle activation patterns or synergy patterns, it has to be reminded that muscle synergies are centrally organized patterns of muscle activity and that the *application of synergy patterns* is one of the concepts for postural control *(Horak et al. 1997)*. In contrast to earlier views, these muscle synergies are flexible and adaptable to characteristics of the initial condition or the perturbation. The second concept of postural control is the application of *movement strategies (Horak et al. 1997)*. Those movement or postural strategies are thought to be distinguishable according to the aim of the CNS, i.e. by what the CNS is attempting to control, and are characterized by their kinematic and kinetic pattern and their diverse muscle synergies *(Horak et al. 1997)*. For example, the main aim for the human system during unperturbed *standing* is the maintenance of the center of gravity within the base of support *(Winter et al. 1990; Winter 1995)*.

Recovery strategies have been classified according to the largest observable movements *(ankle, hip and stepping strategy)*. Since unexpected perturbations lead to a sudden acceleration of the center of mass, the application of compensatory reactions after sudden, unpredictable balance perturbations is essential for the *deceleration* of the center of mass *(Maki & McIlroy 2006)*. The motion of the center of mass can be slowed down by the generation of muscle torque at the ankle, knee, hip or other joints. The *"ankle strategy"* involves a shift of the body´s center of gravity by the rotation of the body around the ankle joints. The *"hip strategy"* implies weight shifts at the hip, including repositioning of the center of mass by flexing or extending the hips. These two strategies pertain to the "fixed-support" strategies *(Maki & McIlroy 1997)*. Larger perturbations may further necessitate a realignment of the base of support which is called the *"stepping strategy"*. However, the stepping strategy and also grasping movements, which both together compose the *"change-in-support"* strategy, are not only mechanisms of last resort, but have shown to be common reactions to postural perturbations *(Maki & McIlroy 1997)*. Control mechanisms for the stepping or "change-in-support" reactions have to integrate the swing leg selection and spatio-temporal characteristics of the foot trajectory and thereby also to account for the speed of the compensatory step *(Maki & McIlroy 1999)*. From a biomechanical point of view, the control of the center of mass in relation to the base of support, i.e. the control of dynamic stability, is based upon three mechanisms. Those mechanisms are *a) "increase of the base of support" or "moving the center of pressure", b) "counter-rotating segments around the center of mass" and c) "application of external force" (Hof 2007)*. Regarding these underlying mechanisms, ankle and hip strategy may be assigned to the mechanism *"counter-rotating segments around the center of mass"* and the stepping strategy can be allocated in the mechanism *"increase of base of support"*.

During *walking*, the center of gravity may never be within the area of the foot *(Winter et al. 1990)* and therefore the central nervous system has to apply different control strategies. During the double stance phase the center of gravity lies somewhere between the two feet, residing in the base of support of both feet. However, outside of the double stance phase, the position of the center of gravity is not necessarily within the base of support. This means that during walking the body is in a continuous state of falling and recovering *(Winter et al. 1990)*. Regarding perturbations during walking, the control of dynamic stability is both phase-dependent and perturbation-specific *(Nashner 1980; Winter et al. 1990; Tang & Woollacott 1999)*. The impact of *tripping* perturbations, for instance, depends on the timing of the perturbation in relation to the gait cycle. During walking, the timing of the trip stimulus within the swing phase influences the preferred selected strategy. Perturbations in the early swing result mostly in *elevating* strategy recoveries *(Schillings et al. 2000)* whereas the obstructed limb is lifted over the obstacle. Perturbations in the late swing, on the other hand, result in *lowering* strategy recoveries *(Schillings et al. 2000)*. In the lowering strategy, the obstructed limb is placed prior to the obstacle and the contra-lateral limb is lifted over the obstacle *(Eng et al. 1994)*.

Those strategies or mechanisms in response to perturbations during standing and walking are complex and purposeful and not rigidly regulated like reflexes *(Dietz et al. 1987; Horak et al. 1997).* Rather, they are able to adapt to specific situations and may be learned through experience in various environmental contexts *(Horak & Nashner 1986)*. *Figure 2-6* shows the conceptual framework by *Horak et al. (1997)* which assumes that the movement strategies are based on behavioral goals, specific tasks and the environmental context. These general conditions underlie biomechanical and neural constraints and generate the required muscle output according to the prioritization of control variables such as control of the center of mass, head or trunk orientation *(Horak et al. 1997).* Factors, which influence the output of the postural response, are the sensory environment, postural orientation, dynamics of control, cognitive resources as well as experience and practice and the perception of goal and context *(Horak 2006).*

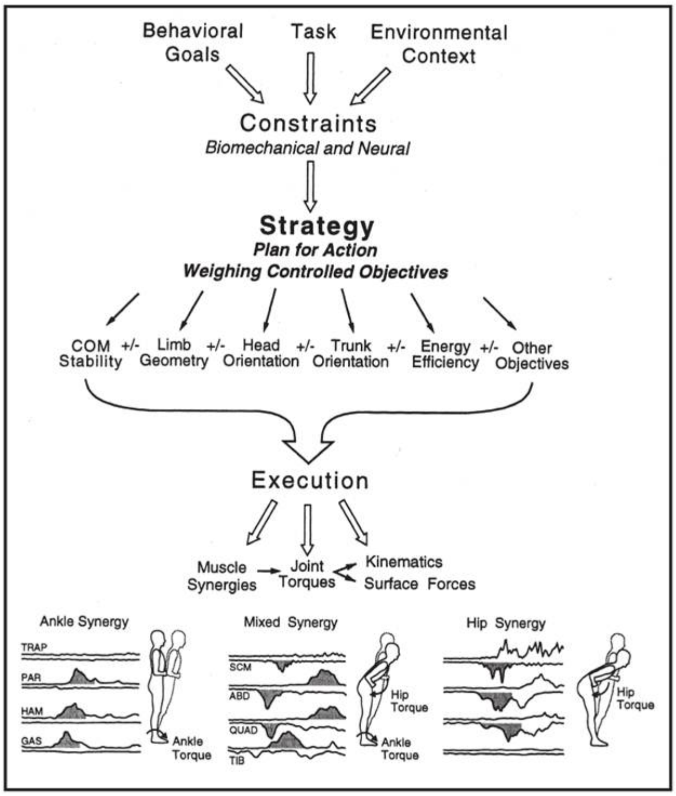

Figure 2-6: *Frame work for the selection and execution of recovery strategies. Strategies are thought to be emergent neural control processes which are based on the behavioral goals, the task and the environmental context and which provide a „plan for action". COM=center of mass, TRAP=trapezius muscle, SCM=sternocleidomastoid muscle, PAR=lumbar paraspinal muscles, ABD=rectus abdominis muscle, HAM=hamstring muscles, QUAD=rectus femoris muscle, GAS=gastrocnemius muscles, TIB=tibialis anterior muscle.*

Reprinted from Horak et al. (1997, p.520) Postural Perturbations: New Insights into the Treatment of Balance Disorders. Phys Ther, 77, 517-533 with permission of the American Physical Therapy Association. This material is copyrighted and any further reproduction or distribution requires written permission from APTA.

Evidence for the effect of different conditions and constraints on the selection and execution of these strategies comes from various studies. The influence of parameters like intention, learning, adaptation and dual tasks on the performance of postural responses *(Quintern et al. 1985; McIlroy & Maki 1995; Burleigh & Horak 1996; Woollacott & Shumway-Cook 2002)* permits, on a behavioral basis, the assumption that the cerebral cortex contributes to the responses. These factors have shown to play a role in the later phases of the postural response *(Horak et al. 1997; McIlroy et al. 1999; Woollacott & Shumway-Cook 2002; Jacobs & Horak 2007)*. Regarding the adaptation to repeated perturbations, for example, it is suggested that large perturbations, which show the development of altered response patterns after the experience of several perturbed trials, may lead to a modification of the cortical control *(Quintern et al. 1985; McIlroy & Maki 1995)*. Feet-in-place responses, however, show no attenuation in consequence to repeated perturbations, explicable by the assumed localization in brainstem neural loops *(Jacobs & Horak 2007)*.

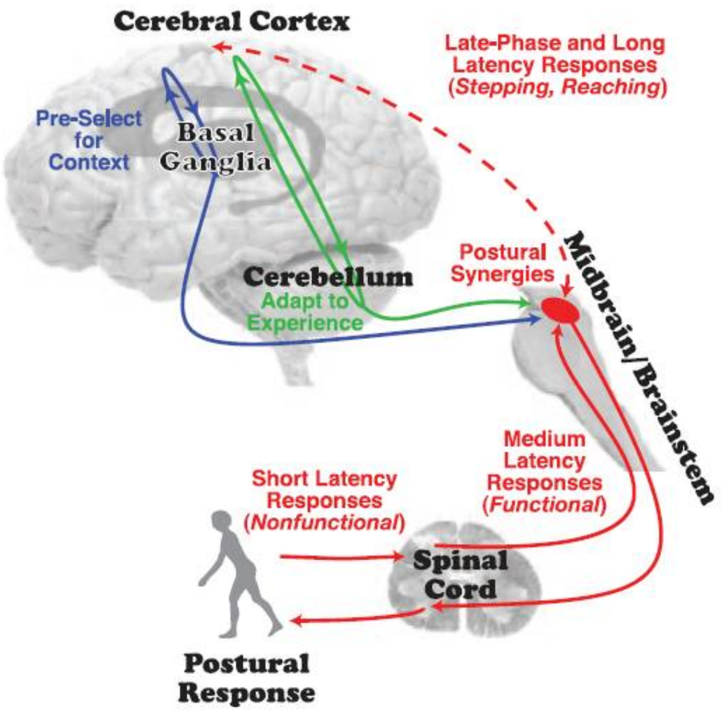

Figure 2-7: *Model of proposed neural pathways which are involved in the control of recovery responses. The contribution of the short-latency activation is quite small, whereas whole body synergies, which include the medium and long latency responses are functionally relevant. Short latency responses can be seen at about 40-50ms and long latency responses at about 120 ms after a stimulus. Reprinted from Jacobs & Horak (2007, p.1341) with permission by Springer Verlag Wien.*

Furthermore, on a neuronal level, the latency of postural responses is longer than that of spinal stretch reflexes but significantly shorter than that of voluntary movements *(Jacobs & Horak 2007)*. Stimulating the motor cortex through single-pulse transcranial magnetic stimuli in the late phase of the postural response revealed an increase in the H-reflex of the soleus muscle and in the magnitude of the postural response *(Taube et al. 2006)*. This indicates an increased cortical excitability in the late phase of the postural response. Even when the extent of cortical control is still not clear, it is accepted, that the probability of central control in the short-latency, medium-latency and long-latency components of postural responses is increased according to the length of the latency *(Taube et al. 2006; Jacobs & Horak 2007)*. *Figure 2-7* shows a recent model of proposed neural pathways which are responsible for short, medium and long latency responses in consequence to perturbations.

According to that model and additional studies, short latency responses are on the level of the spinal cord and provide the first, but simple and quite small postural responses *(Jacobs & Horak 2007)*. It is suggested, however, that the central nervous system may influence even this first reflex response of the muscle due to presynaptic inhibition in the spinal cord *(Dietz et al. 1984; Taube 2012)*. Medium latency and long latency responses following this first, short latency response contribute to whole body synergies which serve to stabilize the body. The initial phase of the postural response seems to be controlled by the spinal cord and the brainstem – later on in the response there is evidence for transcortical reflex pathways, indicating participation of the cerebral cortex *(Christensen et al. 2000; Taube et al. 2006)*. This may imply that compensatory balance reactions like change-in-support responses are characterized by an initial automatic phase and a late phase, in which a contribution of the cerebral cortex and thereby an influence of cognition may be possible *(Norrie et al. 2002; Jacobs & Horak 2007)*. However, the amount of cognitive processing which is required for the adequate performance of response movements depends on the complexity of the postural task and on the capability of the subject´s postural control system *(Horak 2006)*.

In general, the control of static and dynamic stability relies on an accurate internal representation of stability limits. Those stability limits are related to the physical condition of an individual and are based on the ability to execute certain recovery movements or strategies *(Forner-Cordero 2003)*. Therefore, the stability limits are a function of anatomical, physiological and environmental constraints *(Pai et al. 2003)*. It is assumed that the human system controls the stability state in consideration of the motion state of the body center of mass *(i.e. instantaneous position and velocity)* with regard to the base of support *(Pai 2003; Pai et al. 2003; Hof et al. 2008)*. This means, that the relationship of the motion of the center of mass and the base of support together with the underlying individual constraints form the stability limits for each person. The internal CNS representation of the position and motion of the center of mass relative to the base of support has to be accurate to produce adequate control processes *(Maki & McIlroy 1999)*. A possible control strategy of the CNS regarding the control of the center of mass relative to the base of support could be the maximization of the *"stability margins" (Maki & McIlroy 1999; Hof et al. 2005)*. According to this control strategy increasing the step length and the speed of the stepping response may be adequate mechanisms for achieving a better stability during a reactive stepping movement.

2.3.2 Assessment of dynamic stability

There are several possibilities to assess the stability of the human system, but the establishment of a quantitative criterion for the estimation of postural control is quite difficult. The informative value and the validity of the assessment methods vary and only some of the methods will be mentioned here. Since most falls occur during dynamic situations *(Tinetti et al. 1988; Rubenstein 2006)* and conclusions from investigations about static postural control are not directly transferable to dynamic stability control *(Mackey & Robinovitch 2005; Owings et al. 2010)*, the literature review concentrates on studies about dynamic stability control.

2.3.2.1 Motor performance tests

In the clinical field measures of balance and lower extremity function are often performed with standardized tests such as the Functional Reach Test, Timed up & Go or the Performance Oriented Mobility Scale. Sensitivity and specificity of these tests for the prediction of fall rates depend on the type of test. The Berg Balance Scale, for instance, has shown to have a quite poor capability to predict falls *(Bogle Thorbahn & Newton 1996; Boulgarides et al. 2003)*, whereas the Timed Up & Go Test and the Functional Gait Assessment revealed good sensitivity and specificity *(Shumway-Cook et al. 2000; Wrisley & Kumar 2010)*. Those motor performance tests are easy to perform and have shown to be reliable and valid, but they also have limited value. For instance, the association with patient provided history of falls is quite low *(Caterino et al. 2009)* and the ceiling effect within the tests limits the information for healthy old adults. Further, reactive responses to perturbations are mostly not measured by those tests.

2.3.2.2 Assessment of variability of gait by spatio-temporal parameters

Spatio-temporal parameters and kinematic data of the gait cycle have often been used to analyze the gait whereas the linear variability of temporal measures of swing and stance is able to distinguish between fallers and non-fallers *(Hamacher et al. 2011)*. Reliable measures for the assessment of the linear variability are standard deviation or coefficient of variance of swing, stride and stance times. Furthermore, variability of spatial parameters such as step width or step length has also been applied in previous studies *(Maki 1997; Brach et al. 2005)*. The variability of gait has been defined as the variance of gait parameters around the mean, whereas earlier biomechanical studies traditionally assumed that walking variability can be equated with stability. In contrast to that, the stability of gait currently is defined as the resilience or the capacity of the neuromuscular system to restore or maintain function despite (infinitesimal) perturbations during walking. In addition, the statistical measures of variability do not account for the spatio-temporal structure of time series data *(Dingwell & Cusumano 2000; England & Granata 2007)*.

2.3.2.3 Assessment of variability of gait by non-linear methods

An alternative approach to assess gait stability is the application of nonlinear methods. However, measurements of stride-to-stride variability with linear methods and analysis of dynamic stability with nonlinear methods reveal different aspects of locomotor behavior. When assessing gait variability from the differences among strides, each stride is considered as an (statistical) independent event. In this case, previous states of the locomotor system as well as feedback loops in the motor control of walking are not taken into account *(Buzzi & Ulrich 2004)*. In contrast to that, the correlations between consecutive gait cycles and non-linear dependencies, i.e. the dynamic nature of the human gait, are considered by the use of nonlinear methods *(Terrier & Dériaz 2011)*. A modified random-walk analysis *(Detrended Fluctuation Analysis)* and a spectral analysis can reveal long-range, power-law correlations *(Hausdorff et al. 1995; Hausdorff 2007)*. Variability measures and a non-stationarity index are thought to provide information about the regulation of locomotor function *(Hausdorff 2007)*. This analysis method, the stride interval dynamics, assumes an underlying stochastic process and represents the statistical persistence of the temporal fluctuations within an individual´s stride interval time series *(Chang et al. 2009).* This approach attempts to quantify the amount of correlation within a data set and has shown to be able to differentiate between fallers and non-fallers *(Herman et al. 2005)*.

Orbital stability can be measured by the calculation of *maximum Floquet multipliers*. This method quantifies the tendency of the system´s states to return to the periodic limit cycle orbit after small perturbations, i.e. the magnitude of changes of small perturbations after one subsequent stride *(Dingwell & Kang 2007)*. Floquet multipliers are able to distinguish between fallers and non-fallers *(Granata & Lockhart 2008)* as well as between young and old subjects *(Kang & Dingwell 2009)* and showed to reflect reduced stability *(McAndrew et al. 2011)*. However, the method of maximum Floquet multipliers shows inconsistent results towards sensitivity to reduced stability *(Bruijn et al. 2011)* and further assumes purely periodic motion.

The *system maximal Lyapunov exponent* characterizes the average rate of *divergence* in pseudo-periodic processes and can also be used to quantify the sensitivity of a dynamical system to small perturbations – and therefore to quantify the *local dynamic* stability *(Rosenstein et al. 1993; Dingwell & Cusumano 2000; Dingwell & Kang 2007).* In contrast to the Floquet analysis, the full temporal and spatial structure of all trajectories is used within this approach *(Dingwell & Kang 2007)*. Further, in contrast to the Detrended Fluctuation Analysis, a deterministic process is assumed. Local dynamic stability is the ability to attenuate effects of local walking perturbation and is not related to gait variability *(Dingwell et al. 2001)*. The Lyapunov exponent measures the time-dependent rate of kinematic expansion *(Dingwell & Cusumano 2000; England & Granata 2007)* and may be used for the prediction of the probability of falls and the detection of gait instabilities *(Bruijn et al. 2010; Van Schooten et al. 2011)*.

All those non-linear methods, used to assess within-subject stride-to-stride changes in gait (e.g. per video-based marker systems, accelerometers, goniometers), must be able to record a relatively large number of strides *(Hausdorff 2007)*. Specific gait modifications can only be detected by using a substantial number of strides *(Bruijn et al. 2009)*.

2.3.2.4 Assessment of dynamic stability in consideration of the center of mass relative to the base of support

An established concept for the control of balance in *standing* is based on the position of the center of gravity in relation to the base of support. If the center of mass position is located inside the boundaries of base of support, balance should be maintained *(Patla et al. 1990; Winter et al. 1990)*. An expansion of this method is the *Time-to-contact* approach which combines information about the instantaneous position of the center of mass and the boundaries of the base of support to predict a future time point at which the center of mass will pass the boundary of base of support *(Haddad et al. 2006; Hasson et al. 2008)*. By the consideration of the horizontal velocity of the center of mass, the destiny of the center of mass is taken into account in relation to the boundaries of base of support. Similarly, the *feasible stability region* approach considers the horizontal velocity of the center of mass *(Pai & Patton 1997, fig. 2-8)*. This approach is able to quantify the dynamic stability even in dynamic situations such as walking. The equations of motion for this concept are based on an inverted pendulum and either a two-segment sagittal model *(Pai & Patton 1997)* or four-segment model each with a foot segment *(Iqbal & Pai 2000)*. The feasible stability region encloses all possible combinations of center of mass velocity and position *(Yang et al. 2009)*.

The *extrapolated center-of-mass concept* of *Hof et al. (2005)* – and similarly the *concept of the velocity stability margin* by *Maki & McIlroy (1999)* – is based on the feasible stability region approach. However, instead of defining the feasible stability region by computer simulation and an optimization routine, the approach of Hof et al. (2005) applies simple mechanical reasoning to quantify the dynamic stability. The extrapolated center of mass is similarly based on the inverted pendulum model and includes the horizontal velocity of the center of mass. The defined margin of stability is used as a quantity for the dynamic stability, whereas positive values of this parameter indicate a stable position of the human system and negative values an instable position. This approach is supposed to be also applicable in disturbed movements. Measurement of dynamic gait stability in consequence to an unannounced slip showed to be more sensitive for the prediction of fall outcome than most clinical tests *(Bhatt et al. 2011)*.

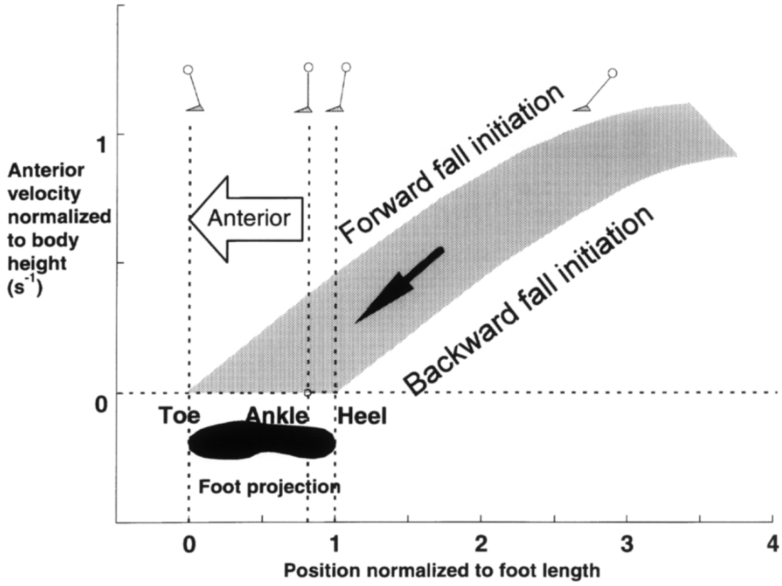

Figure 2-8: *Feasible stability region concept based on center of mass velocity-position limits. The shaded band represents the feasible horizontal center of mass velocity-position region within which an anterior movement of a simple pendulum, connected to a stationary base of support, is thought to be stable. Beyond that range, forward or backward falls would be initiated. Reprinted from Pai & Patton (1997, p.350) with permission by Pergamon.*

2.3.2.5 Comparison of different assessment approaches to quantify dynamic stability

Altogether, each of those assessment approaches has different advantages and disadvantages. The adequate method has to be chosen in consideration of the respective aims of the study. Motor performance tests are easy to perform and to analyze and are especially suited for clinical settings. Those tests, however, are of limited value for old adults which are high functioning – even when they are at risk of falls. With regard to gait stability, diverse approaches try to make statements about the fall risk looking at walking patterns and to predict future falls. The quantification of the variability of spatio-temporal parameters can be used to differentiate between fallers and non-fallers, but is not able to draw conclusions about the regulation of locomotor function. When interested in the locomotor regulation, non-linear measures can reveal dependencies and correlations, but they also require a large amount of steps. Considering postural responses to perturbations, the aforementioned approaches are somewhat limited and are not able to supply information about the stability state direct afterwards a perturbation. Assessment approaches which consider the center of mass behavior relative to the base of support have shown to be auspicious and to be able to quantify the dynamic stability state.

2.3.3 Gait in elderly

As aforementioned, gait of humans has to deal with two disadvantages: the bipedal locomotor pattern, which consists of single-limb support periods and the body structure, which implicates a high center of mass. The rhythmic pattern of human gait is established in the central pattern generators in the spinal cord. Supraspinal control, however, is still important in the human being. The centers in the spinal cord are controlled by locomotor regions in the brain and interact with sensory systems. Premotor and motor areas of the frontal cortex control the gait – those areas are connected with the basal ganglia and the cerebellum which then control the spinal level. Recent studies have shown, that gait is even an attention-demanding task, requiring a higher control of executive processing and memory during challenging tasks and especially in the elderly *(Yogev-Seligman et al. 2008)*. The attentional demand of gait for the elderly in comparison to young adults has been investigated by using dual tasks. The performance of cognitive tasks while walking showed to increase reaction times of the cognitive task or to reduce the gait speed relatively more for older adults compared to young adults *(Lajoie et al. 1996; Lindenberger et al. 2000; Hollman et al. 2006)*. Gait stability, however, seems to be affected by dual tasks in some studies *(Lindenberger et al. 2000; Dubost et al. 2006)*, but not in others *(Li et al. 2001; Springer et al. 2006)*.

Generally, there are some alterations in the coordination pattern of elderly individuals which may be consequences of the functional decline in the various systems. Those changes in the walking pattern may have an influence on the dynamic stability and on the ability to cope with perturbations. Regarding the walking pattern, older adults show shorter step lengths and a broader base of walking, resulting in an increased stance time and double support time *(Winter et al. 1990; Lockhart et al. 2005; Sturnieks et al. 2008, fig. 2-9)*. Elderly adults are thought to prefer shorter step lengths to reduce the ground reaction force which in turn decreases the balance challenges to the upright stability of the upper body in the sagittal plane *(Woollacott & Tang 1997).* Regarding the step width, hip abductors are thought to play an important role in the regulation *(Winter 1995)*. Extreme step width variability (either too high or too low) in individuals who walk at a normal gait speed is associated with falls in the past year *(Brach et al. 2005)*. This indicates a specific range of variability for the step width – beyond that range hip abductors possibly are not able to regulate the step width adequately according to environmental hazards.

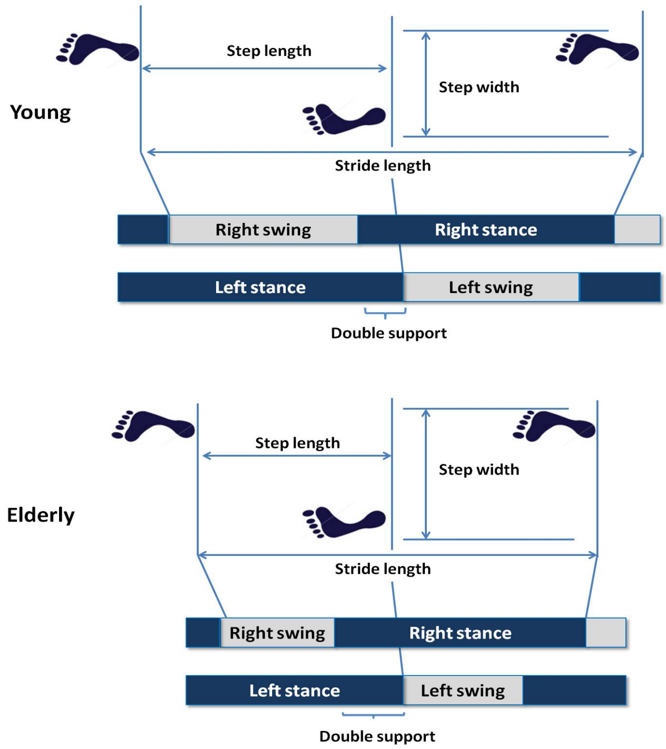

Figure 2-9: *Age-related differences in step length, step width, double support time, stance and swing phases during walking. Elderly adults show shorter step lengths and a broader walking base (Adapted from Chao 1986 with kind permission of Springer Science + Business Media and reprinted from Spirduso et al., p.171 with permission from Human Kinetics.)*

Corresponding to the smaller step length, gait velocity is significantly reduced with advanced age *(Woo et al. 1995; Prince et al. 1997)*. Several studies support the assumption that there is a physiological basis for the reduced velocity *(De Vita & Hortobágyi 2000; McGibbon 2003)*. However, even at the same walking speed, fall-prone elderly individuals reveal a less dynamic stability than healthy older or young adults, measured by methods of Poincare analyses *(Granata & Lockhart 2008)*. In order to increase their gait velocity, older adults prefer to increase their stride frequency whereas young adults increase their stride length *(Larish et al. 1988)*. Stride length differences therefore are accentuated as the walking speed is increased *(Larish et al. 1988)*. The preference of increased stride frequency may be chosen by the elderly because of an impaired flexibility or the avoidance of decreased double stance time *(Larish et al. 1988)*. Changes in walking speed and stride parameters with aging are explained by the compensation for muscle weakness *(Frontera et al. 1991; Gibbs et al. 1996)*, coping with balance impairments *(Schultz 1995)* and increased variability *(Danion et al. 1993)* and a reduction of energy cost *(Holt et al. 1995; Minetti et al. 1995)*. Several changes in the walking pattern like the decreased gait variability or the increased double stance phase are associated with fall prevention behavior *(Barak et al. 2006)*.

Regarding the kinematic profile of gait in elderly, elderly persons show a lower dynamic range of motion in the ankle and knee *(Winter 1991; Prince et al. 1997)*. Investigations about the dynamic range of motion in the hip are contradictory – some studies show a larger *(Winter 1991)* and some show a lower range of motion *(Kerrigan et al. 1998)*. However, the exerted torque and power in the lower extremity reveal an age-related redistribution among the joints. With ageing, an increased output is generated by the hip musculature whereas the musculature of the more distal joints produces less output, especially in the ankle joint *(De Vita & Hortobágyi 2000; McGibbon 2003)*. This redistribution is ascribed to a compensating function of the hip which is required because of decreased strength and function of distal muscles, observable also in a decreased push-off power of the ankle during gait *(Winter et al. 1990; De Vita & Hortobágyi 2000).* The reason for the age-related gait adaptation may be the decreased muscle strength due to a loss of motor neurons, muscle fibers and aerobic capacity *(Woo et al. 1995; Prince et al. 1997)*. During higher walking speeds, elderly fallers showed increased hip flexion in the swing phase which may help to achieve a specific step length despite the reduced push-off which may possibly minimize propulsive forces *(Barak et al. 2006)*.

When walking on irregular surface, older adults revealed a more conservative strategy compared to young – gait changes of the elderly (deceleration of gait speed, broadened step width and increased step time) were more pronounced in comparison to young individuals *(Menz et al. 2003; Thies et al. 2005)*. This conservative strategy is probably chosen to increase stability. Furthermore, during walking on a multi-surface terrain, older adults demonstrated also greater medio-lateral trunk acceleration and trunk pitch as well as roll variability compared to young adults *(Marigold & Patla 2008)*. These characteristics increase the fall risk because of the high velocity of the center of mass towards the limits of the base of support. Lateral instability, also during standing, seems to be an indicator for fall risk *(Lord et al. 1999)*.

Regarding the differences in gait between old and young individuals, the variability of specific *spatio-temporal parameters* such as the standard deviation of step width or stride time or the coefficient of variation of stride velocity have been identified as sensitive factors for the differentiation between the two age groups *(Barak et al. 2006; Hamacher et al. 2011)*. The elderly show a greater variability in several gait parameters and this is attributed to decreased leg strength and passive ranges of motion *(Kang & Dingwell 2008*). Further, nonlinear measures of the variability of stride time also allowed differentiation between older and young adults *(Hausdorff et al. 1997: Hamacher et al. 2011)*. More important with respect to the fall risk, however, is the differentiation between fallers and non-fallers. In this case, linear variability (standard deviation and coefficient of variation) in stride, swing and stance time can be used to distinguish between fallers and non-fallers. Additionally, nonlinear measures revealed fall-related differences in the variability of minimum foot clearance and stride trajectory *(Barrett et al. 2010; Hamacher et al. 2011)*. In general, gait instability (increased linear variability or altered nonlinear variability of foot trajectories) is higher in elderly as well as in fall-prone adults compared to young *(Hamacher et al. 2011)*. The variability of kinematic variables has shown to be an important gait risk factor for falls and therefore should be considered when assessing the gait of elderly *(Barak et al. 2006)*. An increased variability in the stride symmetry may have an influence on the likelihood of obstacle contact and therefore on the risk of falls.

However, it is not only important to identify and assess the differences in gait in elderly and young as well as in fallers and non-fallers – it is even more important to identify the specific causes for the falls. When studying postural control and mobility, the interaction between the individual, the task and the environment has to be considered *(Shumway-Cook & Woollacott 2007)*. This means that the study of balance control during walking in elderly should take into account the nature of the tasks (walking) and the environment that puts the elderly at a higher risk of falling *(Woollacott & Tang 1997)*. According to *Rubenstein (2006)*, the most frequently cited cause of falls in elderly is “accidental” or environment related. This cannot only be ascribed to identifiable environmental hazards, but it is also a consequence of the above-mentioned age-related gait changes. Therefore, the *interaction* of environmental hazards and an increased individual susceptibility to hazards, for example because of age effects like impaired coordination, a more variable gait pattern and a decreased muscle strength, leads to an increase in the fall risk with age *(Rubenstein 2006)*.

2.3.4 Postural adjustment strategies in the regulation of perturbation in older adults

According to *Kovacs (2005)* there are two main reasons for falls in elderly: the first reason refers to a failure to detect a misplacement of the body´s center of mass because of sensory deficits. The second reason refers to an inappropriate response to a perturbation of the center of mass. This inappropriate response may be the consequence of decreased strength or an inability to perform the correct motor response. Both reasons play a role especially in the context of environmental hazards.

2.3.4.1 Proactive adjustments to environmental hazards in old age

With regard to *proactive* handling of environmental hazards, older adults show preserved but modified obstacle avoidance capabilities in comparison to young adults *(Kovacs 2005)*. Stepping over obstacles of low height seems to be no problem, but stepping over obstacles of greater height confronts elderly with more problems *(Patla et al. 1991)*. During walking over obstacles, elderly show an increased height of the swing leg above the obstacle (increased elevation) whereas the swing velocity is decreased *(Chen et al. 1991; Patla et al. 1991)*. Older adults further use shorter step lengths and shorter obstacle-heel strike distances which is associated with the use of more conservative stepping strategies *(Chen et al. 1991)*. The principle of conservatism can be seen in cautious, carefully performed movements and is used to reduce the potential risk. It is assumed that the application of the conservative stepping strategy may be used to achieve a decrease in the momentum and movement speed and accordingly to achieve an increase in the stability margin (i.e. more stable position) *(Kovacs 2005)*. However, although potential toe-contact was minimized by the application of this strategy, the shortened step length increased the risk of stepping on the obstacle with the heel *(Chen et al. 1991, fig. 2-10)*.

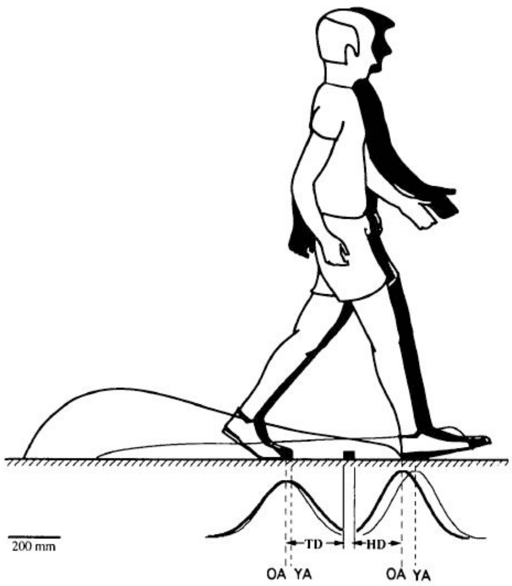

Figure 2-10: *Age-dependent strategy for crossing an obstacle. Distribution of heel distance (HD) and toe distance (TD) are shown for old (OA, light) and young adults (YA, dark). Reprinted from Chen et al. (1991, p.M201) with permission by The Gerontological Society of America.*

During time-constrained conditions, i.e. a short available time frame for the movement adjustment, older adults further showed a more frequent contact of the obstacles compared to young *(Galna et al. 2009)*. When older adults were asked to match their stride (lengthen or shorten) to the position of a cue during walking, they showed to be able to perform the task similarly like the young adults when the cue was given two steps in advance, but revealed difficulties when the cue was given only one step in advance *(Patla 1993 in Woollacott & Tang 1997)*. Further, old adults showed more difficulties in adjusting the step length when they had to *shorten* the step to match the cue *(Patla 1993 in Woollacott & Tang 1997)*. Approaching an obstacle, elderly additionally showed a lower rate of success in avoiding the obstacle during time-critical conditions compared to young – indicating similarly the need for a longer period of time to implement gait modifications *(Chen et al. 1994)*. However, having a relatively long preparation time in advance to environmental hazards, the approach and accommodation to known surface height changes challenges the gait of elderly still more than the gait of young adults *(Lythgo et al. 2007)*. This is apparent in a reduction of step velocity and a more cautious gait. The preparation time of several steps in advance in this study led to earlier and larger step adjustments and a more frequent short step crossing strategy for the elderly compared to young adults *(Lythgo et al. 2007)*. Those adjustments and correspondent reduced footfall variability may be indications for a greater proactive control or cautious behavior for the elderly *(Lythgo et al. 2007)*. Proactive balance control therefore demands higher attention for the elderly compared to young adults – at least when performing complex movement tasks – leading often to worse proactive adjustments during dual tasks *(Chen et al. 1996)*. However, dual tasks during perturbed gait also revealed a better predictive behavior compared to single task performance for elderly, attributable to increased arousal *(Bohm et al. 2012)*. When challenged with multiple obstacles, high risk older adults choose a potentially hazardous gaze strategy, namely a prioritization of the planning of future stepping actions over the accurate execution of ongoing steps *(Chapman & Hollands 2007)*. The premature gaze transfer was associated with a decline in stepping accuracy and precision *(Chapman & Hollands 2007)*.

2.3.4.2 Reactive adjustments in consequence to perturbations in old age

In comparison to predictive adjustments, coping with sudden and unpredictable perturbations demands fast and relatively large adjustments of the postural responses. This may provide higher challenges for the older compared to the young adults.

In response to perturbations during standing, older adults show a shift towards a use of the *"hip strategy"* rather than using an *"ankle strategy"* which is preferred by young adults *(Shumway-Cook & Woollacott 2007)*. This is attributed to ankle weakness or loss of peripheral sensory function. Older adults further tend to initiate a stepping response at lower levels of instability than young adults *(Jensen et al. 2001; Sturnieks et al. 2012)* and rely more frequently on multiple steps to restore balance *(McIlroy & Maki 1996; Pavol et al. 2002)*. It is assumed that the increased likelihood of stepping responses for the elderly is the consequence of a more undamped acceleration of the head for the elderly what leads to a perceived higher threat compared to young *(Jensen et al. 2001)*.

More demanding tasks are frequently performed by the use of tether release experiments which simulate trips. At tether-release studies, participants experience a simulated forward-fall and have to prevent a complete loss of balance by taking one or more steps. In comparison to young adults, older individuals are less able to recover balance from larger forward lean angles *(Wojcik et al. 1999; Hsiao-Wecksler 2008)*. Regarding similar forward lean angles for young and old, younger adults are able to create a higher margin of stability at touchdown – independent of perturbation intensity *(Karamanidis et al. 2008; Mademli et al. 2008)*. The higher margin of stability is achieved by the application of large and quick recovery steps for the young *(Karamanidis et al. 2008; Mademli et al. 2008)*. The recovery deficit of the elderly may therefore be attributed to slower peak joint velocities in hip flexion, knee flexion and extension and ankle plantarflexion for the old compared to young adults *(Madigan & Lloyd 2005; Arampatzis et al. 2011)*. Both, neural and muscular age-related effects could underlie the slower leg movement – differences in the timing of the muscle activation during this task, however, indicate that neural factors play an important role *(Thelen et al. 2000)*.

Lateral stability of older individuals was investigated in response to perturbations induced by a motor-driven waist-pull system. Older fallers demonstrated a greater sideways body motion towards the stepping side at first-step contact as well as a more laterally directed foot placement compared to non-fallers (old and young) *(Rogers et al. 2001)*. The anticipatory postural adjustments for the minimization of lateral instability, however, were similar for the older fallers, older non-fallers and young adults *(Rogers et al. 2001)*. The wider, more lateral step for the older fallers is thought to be applied either by compensating the instability between take off and touchdown of the foot or because of preplanning to compensate for the lateral instability – possibly driven by anxiety about falling *(Rogers & Mille 2003)*. In response to lateral perturbations of stance, old adults showed a similar recovery behavior compared to younger adults: both used rather a sequence of side steps than crossovers *(Maki et al. 2000)*. However, older adults were more likely to use arm reactions or to take extra steps and they showed a higher rate of collisions between swing foot and stance limb *(Maki et al. 2000)*.

In consequence to *slips,* older adults are over twice as likely to fall as young adults *(Pavol et al. 2002; Pai et al. 2010)*. Strategies for regaining balance after a slip shift from the *"hip strategy"* to the *"stepping strategy"* with ageing *(Rubenstein 2006)*. With age, not only the response to slips is modified, but also the susceptibility to slips. This can be seen in an increased heel velocity at heel contact during walking for the elderly adults as well as a slower transitional acceleration of the center of mass which may influence the friction demand *(Winter 1991; Lockhart et al. 2003)*. When experiencing a slip during walking, young adults regulated their forward progression velocity by an increase of the stride duration and stride length *(Tang & Woollacott 1999)*. In contrast, old adults, in contrast, seem to be unable to use a longer stride length in order to increase their base of support *(Tang & Woollacott 1999)*. The application of a shorter stride length was assumed to reflect a more conservative balance strategy in balance threatening conditions *(Tang & Woollacott 1998, 1999)*. In response to slipping perturbations, older adults show a different muscle activation pattern and a limited flexibility for modulating the magnitude of the postural responses compared to young adults. However, older adults preserve the ability to modulate their muscular responses depending on the magnitude and timing of the perturbation with ageing *(Tang & Woollacott 1999)*. The increased fall rate of the elderly after a slip is further attributed to an age-related decline in the ability to apply self-selected recovery responses *(Pavol et al. 2002)*.

Tripping perturbations often reveal diverse preferred recovery strategies for young and older adults. In mid-swing, the response to a perturbation is not as clearly assigned as in early- or late swing phase. As a consequence of perturbations during mid-swing, elderly individuals prefer more often a lowering strategy in comparison to younger individuals *(Pavol et al. 2001; Pijnappels et al. 2005)*. It is assumed that reduced response and movement time together with insufficient recovery limb strength deters elderly from the required large recovery step in an elevating strategy and leads to the preferred lowering strategy *(Roos et al. 2010)*. During the early-swing elevating strategy, elderly adults show smaller responses in the ipsilateral upper-leg muscles, leading to shortened step distances and durations compared to young adults *(Schillings* *et al. 2005)*. Further, reactive stepping adjustments in elderly in consequence to an unpredictable shift of a visual target revealed a delayed modification of the forces of the supporting foot and of the stepping foot trajectory *(Tseng et al. 2009)*. Delayed response selection and planning processes accordingly may contribute to the higher susceptibility to falls for the old adults after slips or other perturbations *(Lockhart 2008; Tseng et al. 2009)*. In conclusion, in consequence to unexpected perturbations like obstacles, slipping or tripping, elderly individuals show a decreased *reactive* recovery performance in comparison to young adults *(Tang & Woollacott 1998; Wojcik et al. 1999; Pijnappels et al. 2005)*. Regarding recovery responses, we conclude that not only the rapidness of the response is critical for the handling of perturbations, but also the execution of effective modifications and adequate recovery strategies.

The selection and execution of movement strategies, however, depends on central processes. Dual-task paradigms such as executing cognitive tasks while performing postural tasks revealed altered attentional requirements for the older adults *(Maki et al. 2001; Woollacott & Shumway-Cook 2002)*. Attentional demands for recovery responses are assumed to build a hierarchy in the elderly – feet-in-place movement strategies seem to require lower attentional demands, whereas compensatory stepping responses require higher attention *(Brown et al. 1999)*. Performing the stepping response further revealed higher attention demands for the elderly in comparison to young adults *(Brown et al. 1999)* – indicating either less attention capacity or a greater need for attention during postural recovery for the elderly *(Bishop et al. 2010)*. Older adults, however, favor stepping as a recovery strategy and since stepping seems to be more attentionally demanding for the elderly, the use of this stepping strategy may promote postural instability *(Brown et al. 1999)*. The preference toward stepping compared to other possible postural responses may be the consequence of a deficient muscle recruitment which impairs the use of a feet-in-place strategy *(Rankin et al. 2000)*. It is suggested that the deficient muscle recruitment (reduced EMG activity) in a dual-task condition favors the stepping strategy to ensure a successful balance recovery *(Rankin et al. 2000)*.

2.4 Adaptability

2.4.1 Adaptability of elderly adults

While the rapidness of the postural response concerning motor latencies and contraction velocity is only partly susceptible to training, the appropriate selection and execution of recovery strategies in an adequate time frame may help to regain balance after perturbations. Therefore, with regard to fall incidence, it is important to investigate to what extent recovery strategies of elderly adults can be adjusted after the repeated exposure to perturbations and furthermore if there is a transfer effect to various situations.

2.4.1.1 Sensorimotor adaptation

In the scientific literature, learning new tasks is differentiated mainly in the two sections *"learning of sequences"* and *"sensorimotor adaptation" (Seidler 2006)*. In sequence learning, subjects learn to combine single movement parts to one continuous motion. This ability seems to be not impaired in the older adults *(Howard & Howard 1992; Seidler 2006)*. Only sequences with high complexity revealed impaired sequence learning for the elderly *(Dennis et al. 2003; Howard & Howard 2004)*. The influence of complexity on learning a task refers to the influence of strategic knowledge. Age seems to have minor influence on learning non-declarative tasks, while learning of declarative or strategic tasks, which require conscious processing, is affected *(McNay & Willingham 1998).* This can be attributed to a decline of several cognitive and sensorimotor functions such as reaction time, stimulus discrimination, verbal memory or spatial abilities with age *(Bock & Schneider 2002)*. Regarding neurotransmitter systems, dopamine, which was shown to be important for higher cognitive functions such as working memory *(Arnsten & Li 2005)*, is reduced with age *(Kaasinen & Rinne 2002)*. Since working memory is important for skill acquisition *(Bo & Seidler 2009)*, a decline in this function may contribute to impaired motor learning.

During sensorimotor adaptation, the human system adapts its behavior to changes in the general set-up. The sensorimotor system uses information from several sensory subsystems for the planning and execution of motor actions and has to react adequately to changes. Adjustments to changing conditions in the environment and to changes in the intrinsic properties of the individual human system such as biomechanical properties or the proportions of the body are important to remain functioning during everyday tasks and throughout the life span. This means that subjects modify their movements by sensorimotor adaptation to adjust to changes which come up in either the sensory input or the motor output characteristics *(Seidler 2006)*. Long-term adaptation is seen when an initial decrement of the performance in consequence to changes is followed by a gradual adaptive improvement *(Bock & Schneider 2002)*. In general, motor adaptation is a trial-and-error process of adjusting movement to new demands, based on error feedback *(Bastian 2008)*. Therefore, adaptation is thought to be driven by sensory prediction errors. Those errors can be used to calibrate the internal representations of body dynamics and the environment and further to recalibrate if changes occur in one of those systems *(Bastian 2008)*. Internal representations are important because, if those representations are well calibrated, the human system may decrease the reliance on time-delayed feedback from body sensors and act feedforward *(Bastian 2008)*.

Motor adaptation according to *Martin et al. (1996b)* is defined based on the following three criteria: 1) the movement retains its identity as being a specific pattern, but changes with regard to some parameters; 2) the change occurs with repetition of the behavior and is gradual and continuous; 3) once adapted, subjects cannot retrieve the prior behavior, but show “after-effects” and have to “de-adapt” the behavior with gradual and continuous practice back to the prior state *(Martin et al. 1996b)*. The so-called *“after-effect”* is the sustained modification of a specific movement after the adaptation to a stimulus, which is still observable after the cessation of the impact of the stimulus. Storage of the new learned pattern can be observed by analyzing after-effects. Further, the persistence of the improved performance, thus, the retention of the learned behavior, can be measured by a retention test some time after the performance.

The process of sensorimotor adaptation, following *Richards et al. (2007)*, is comprised of two distinct control processes: strategic perceptualmotor control and adaptive spatial alignment. Strategic control occurs early in the adaptation process when subjects understand on a conscious level what they have to do to correct errors.

Strategic control is the process of providing (cognitive) on-line adjustments to diminish consequences of deficient motor outputs which are not appropriate for an altered sensory environment *(Richards et al. 2007)*. This kind of control, however, is task-specific and does not generalize to other contexts. Adaptive spatial alignment on the other hand takes place when the exposure to the sensory discordance lasts for some time and new strategic sensorimotor coordination patterns are reinforced. Those patterns are reinforced until they become more automatic and therefore adaptive *(Redding & Wallace 1996; Richards et al. 2007)*. The aforementioned after-effects therefore are measures of adaptive spatial realignment, direct effects, though, are measures of strategic control *(Redding & Wallace 2002)*.

Sensorimotor adaptation often is investigated by the use of split-belt treadmills. A recent study found differences in the adaptation and deadaptation rates between conscous corrections and distractions *(Malone & Bastian 2010)*. Conscious corrections during the adaptation phase accelerated the process but also lead to faster deadaptation in contrast to a distracted adaptation process. This indicates the importance of underlying processes and further, the importance of learning conditions *(Malone & Bastian 2010)*. It is assumed that, with regard to rehabilitation strategies, patients should be put in situations which per se drive the learning of new patterns without having to use voluntary effort *(Torres-Oviedo et al. 2011)*. In addition, the results of this study suggest a more flexible and accessible adaptation of spatial locomotor control than temporal control *(Malone & Bastian 2010)*. Different neural structures are thought to be involved in adapting spatial and temporal control *(Torres-Oviedo & Bastian 2010)*. Furthermore, learning conditions also showed to be important with regard to the transfer or generalization effect *(Torres-Oviedo & Bastian 2010).* Removing vision, for instance, increased the transfer effect from treadmill to overground walking – associated with either a change of the person´s perception of the source of error during adaptation or with an upweighting of proprioceptive information *(Torres-Oviedo & Bastian 2010; Torres-Oviedo et al. 2011)*.

2.4.1.2 Sensorimotor adaptation of elderly

With respect to the adaptability of older adults, the acute impairment of sensory subsystems has shown to provide greater difficulties for the older adults and revealed constraints in their adaptation potential to modifications of the available sensory information *(Woollacott et al. 1986; Camicioli et al. 1997)*. Reduced sensory input during standing has shown to lead to a diminished adaptation for the elderly adults, obvious particularly in the oldest old who repeatedly loose balance *(Judge et al. 1995)*. Tracking tests, which also can be assigned to sensorimotor adaptation tasks, have shown higher tracking errors and increased tracking-error variability for the elderly compared to young adults during several repetitions *(Bock & Schneider 2001)*. However, both age-groups performed these tracking tests with similar adaptive improvements *(Bock & Schneider 2001).* These results were also found in other studies *(Fernandez-Ruiz et al. 2000; Roller et al. 2002; Buch et al. 2003)*. Further studies yet found an age-related decline in adaptive improvement *(McNay & Willingham 1998; Buch et al. 2003; Seidler 2006)*, not only according to the type of task (strategic vs. non-strategic) but also regarding to the temporal exposure of changes (gradual vs. sudden distortions). Gradual distortions led to similar adaptations for older compared to young adults whereas sudden changes resulted in an age-related reduced adaptation level *(Buch et al. 2003)*. Independent of that, most studies show no age-related decay of after-effects as an indication for adaptive recalibration *(McNay & Willingham 1998; Buch et al. 2003)* or are even larger in older than in young adults *(Fernandez-Ruiz et al. 2000)*.

Although older adults have shown to be able to learn new skills and to relearn old they learn these skills at a slower rate than young adults *(Spirduso et al. 1995)* and their observed improvements with practice do not always achieve the level of young adults *(Woollacott 1993; Judge et al. 1995)*. Aging with regard to motor learning is assumed to be accompanied with possible deficits in adaptation, which have been subscribed mainly to impaired strategic control, while maintaining the ability to recalibrate *(Bock 2005)*. The observed decline in skill acquisition may be linked with structural changes of the brain with age *(Seidler 2006)* as, for example, volume loss in the caudate and cerebellum *(Raz et al. 1992, 2005)* since the cerebellum has been shown to contribute to sensorimotor adaptation *(Martin et al. 1996a; Imamizu et al. 2000; Morton & Bastian 2006).*

Therefore, it is reasonable to discuss the preservation or decline of the adaptation potential regarding dynamic stability for the elderly. Regarding locomotion, it is evident that the walking pattern has to adapt flexible to different environments and constraints.

The decreased adaptation performance to some tasks for the elderly and moreover age-related sensorimotor deteriorations, which may impair the acquirement of sensory information, may affect the ability of older adults to adapt their locomotor pattern to changes and perturbations. The age-related decline in functional capacity, i.e. reduced strength and power of the lower extremities in comparison to young adults, may further limit the possible adaptation magnitude.

As mentioned before, the postural control relies on predictive and reactive adjustments in response to environmental changes. Predictive and reactive adaptability may be diversely affected by the ageing process because of different underlying mechanisms. Predictive locomotor responses, for instance, depend mainly on supra-spinal structures, especially the cerebellum *(Earhart et al. 2002a; Morton & Bastian 2006)* whereas reactive adjustments are believed to address lower neural centers such as spinal cord or brain stem in the first place *(Morton & Bastian 2006)*. Later on after an unexpected perturbation, the cerebral cortex may additionally contribute to the postural response *(Jacobs & Horak 2007)*.

2.4.2 Proactive Adaptations of elderly adults with regard to perturbations of dynamic stability

Appropriate feed-forward control, based on prior experience and knowledge about the perturbation, may help to reduce the magnitude of the reactive response and therefore to counteract the destabilizing effect. Purely reactive responses in consequence to perturbations are seen only in the first exposure to a perturbation. Following experiences with the same perturbation are characterized by modified timing, magnitude and coordination *(Marigold & Patla 2002; Patla 2003)*. Therefore, even when separating these responses here into different sections, isolating reactive and predictive behavior during different perturbations is difficult. Further, even the observation of perturbations may offer advantages for the performance because of the potential impact of awareness and perceptual knowledge on the handling of the perturbation *(Bhatt & Pai 2008)*. Arousal- and anxiety-mediated changes in the behavior or alterations in the focus can influence the outcome during the adaptation process additionally to predictive locomotor modifications *(Maki & McIlroy 1996; Kelly et al. 2010)*.

The adaptation process according to familiar but potentially dangerous perturbations during walking may take place in relatively few trials in comparison to artificial disturbances like force fields *(Bunday et al. 2006)*. Since the acquirement of a new motor program needs some time, it is unlikely that this is the reason for the rapid adaptive process. The quick adaptation rate in locomotion rather may be possible because of a recalibration of the internal representation of the stability limits *(Wang et al. 2011)*. This means that via trial-and-error practice the central nervous systems learns to reduce the prediction error and recalibrates the internal representation according to the actual conditions *(Bastian 2008; Wang et al. 2011)*.

Repeated exposure to a slip during a *sit-to-stand* task revealed similar adaptation rates for young and older age groups *(Pavol et al. 2002; Pai et al. 2003)*; this can be observed in a similar likelihood of using an initial forward or backward step. The improved stability in consequence to slips is attributed to an enhanced feed-forward control of stability and an adaptive refinement of the internal representation of the stability limits *(Pai et al. 2003*). Proactive adaptations were seen in the position and velocity of the center of mass at seat-off *(Pavol et al. 2004)*. However, despite similar adaptation rates for both age groups, the elderly tended towards using *multiple* steps after the perturbation *(Pavol et al. 2002)* and showed a lower adaptation magnitude *(Pavol et al. 2004)*. Another sit-to-stand protocol similarly showed no age-difference in stability but age-differences in hip height throughout the adaptation phase *(Pai et al. 2010)*. After only one slip exposure, older adults reduced their odds of falls by 8-fold during a sit-to-stand task *(Pai et al. 2010)*. Similarly, namely 7-fold, was the reduction of their odds of falls during walking. This indicates a largely preserved ability to acquire fall-resisting skills *(Pai et al. 2010)*.

Regarding the predictive adaptability of old individuals, only few studies, until now, investigated the adaptation potential of elderly to perturbations during *walking*. When comparing normal gait and gait under expected-slipping conditions, older individuals showed gait modifications already in the preliminary adjustment step, i.e. in the step prior to the expected perturbation *(Lockhart et al. 2007)*. For young adults, it was sufficient to modify their gait within one step prior the perturbation. Modifications were implemented in a reduced heel contact velocity and step length for both age groups when stepping on the slippery surface. Age differences were seen in the step behavior before the contamination and the muscle activation patterns during the adjusted gait, reflecting different strategies. The early parameter adjustments may help the elderly to achieve similar adjustment results on the slipping condition. A more cautious adjustment strategy for the elderly is assumed, accompanied by a higher time-demand *(Lockhart et al. 2007)*. Anyhow, young and old age-groups both modified successfully their gait pattern to be able to traverse the slippery surface.

A study of *Pai et al. (2010)* analyzed the dynamic stability, considering the shortest distance from the center of mass state, and the limits of the feasible region, under slipping walking conditions *(Pai & Iqbal 1999)*. Hereby, old and young adults showed similar adaptation rates as well, observed by the number of balance loss and falls and further by the calculated dynamic stability *(Pai et al. 2010)*. Both age groups showed a steady-state in their adaptation behavior after the exposure to several walking trials under slipping conditions *(Pai et al. 2010)*. Altogether, despite possible aging-related declines in sensorimotor function, elderly adults seem to be able to learn to resist to balance loss and to apply this ability in different contexts *(Pai et al. 2010)*. However, since the dynamic stability can be influenced by several different factors, it remains unclear if young and old adults performed the same mechanisms to achieve similar stability states.

With respect to the literature it is still difficult to draw conclusions about the predictive adaptability of old adults regarding walking. The study by *Lockhart et al. (2007)* did not investigate dynamic stability. Therefore, it is not known if older adults are in a similar dynamic stability state compared to young adults prior to the perturbation and afterwards. Despite similar adjustments, young and older adults applied different adjustment strategies. The consequences of those predictive adjustments are not known and thus we cannot conclude if those adjustments are helpful for both groups equally to deal adequately with the slipping surface. The walking study of *Pai et al. (2010)*, on the other hand, permits no conclusion about the contribution of predictive and reactive adjustments. Since the stability values were analyzed 300 ms after slip onset, both adjustment strategies take part and it is not possible to isolate predictive and reactive adjustments. However, regarding adaptability in age, since different control centers are responsible for predictive and reactive adjustments, older age may diversely affect those.

2.4.3 Reactive Adaptations of elderly adults with regard to perturbations of dynamic stability

Even when predictive behavior is limited because of specific uncertainty about the perturbation, adaptation of the reactive responses is still possible, as earlier investigations have observed *(Horak & Nashner 1986; Horak et al. 1989b)*. Uncertainty in adaptation studies can be produced by unpredictable perturbation magnitude and occurrence time. Experiencing a reslip after a period of non-slipping sit-to-stand tasks revealed a greater incidence of balance loss and lower stability for the elderly compared with young adults *(Pavol et al. 2002; Pai et al. 2010)*. *Pai et al. (2010)* assumed that the short-term retention of adaptation may be affected. However, the previous exposure to several slipping trials reduced the fall incidence for both, young and older individuals, in comparison to the first unexpected slipping task *(Pavol et al. 2002)*, indicating preserved reactive adaptability for the elderly. In consequence to repeated tether-release trials, elderly were also able to improve their stability by enhancing the margin of stability, namely increasing the base of support and the velocity of the recovery step and further by generating a lower velocity for the center of mass *(Carty et al. 2012)*. Repeated exposure to continuous, variable amplitude oscillations of the support surface showed a maintained capacity for the learning of postural responses for the elderly as well *(Van Ooteghem et al. 2009)*. In the time course of learning, older adults performed an improved temporal control of the center of mass and minimized their trunk instability, similarly like young adults. Both age groups yet applied different behaviors, suggesting a more rigid movement strategy for the elderly *(Van Ooteghem et al. 2009)*. The sway energy response in consequence to discrete platform perturbations showed a decrease for the age-group between 66 to 77 years, indicating adaptation – the older age-group (80-102 years), though, was not able to decrease the sway energy accordingly *(Camicoli et al. 1997)*. This could reflect age-related deficits in either the motor output or the integration of sensory output in the elderly old adults.

Repeated exposure to deceleration perturbations during *walking* on a split-belt treadmill revealed a reduced muscular activation pattern, i.e. lower EMG-amplitude of lower extremity muscles, for elderly individuals while maintaining similar muscle latencies compared with the first responses *(Sakai et al. 2008)*. This adaptation to the perturbations led to a smaller body sway, seen in a smaller forward and backward acceleration after experiencing several perturbed trials. It is assumed, that the first reactions to the perturbed walk showed an over-reaction and rather hindered a successful, quick response due to increased co-activation *(Sakai et al. 2008)*. The reactive adaptation to the repeated perturbation therefore has shown to take place in the attenuation of muscle activity and co-contractions *(Sakai et al. 2008)*.

Yet, it still remains an open question how the elderly individuals exactly adapt their reactive responses to unexpected stimuli or perturbations with regard to their dynamic stability state. Additionally, it is not known if the reactive adaptation rate for perturbed walking of elderly is comparable to young adults and if and in what time frame a steady-state is observable.

2.4.4 Interventions for the improvement of dynamic stability in elderly *(long-term adaptations)*

Regarding the adaptability of elderly, it is not only important to analyze the ability to adapt to repeated perturbations which are provided in a short time frame, but it is even more important to investigate the long-term retention of learned behavior and the generalization of specific skills to different situations.

According to *Shumway-Cook et al. (1997)* the goals of balance retraining for older adults are *"(1) to resolve or prevent underlying impairments, (2) to develop effective and efficient task specific sensory and motor strategies and (3) to adapt task-specific strategies so that functional tasks can be performed in changing environmental contexts" (see also Shumway-Cook & Woollacott 1995, p55)*. Till now, several different intervention studies aimed to reduce the fall risk for elderly individuals. There are diverse approaches regarding the same problem and the same goal – some act on the assumption that the main contributor to the increased fall risk is the declined muscle strength, others try to improve balance by exercising balance tasks, flexibility or endurance training. However, recent meta-studies revealed that (progressive) strength training indeed increased strength in older adults, but showed no clear effects on balance tasks *(Skelton et al. 1995; Latham et al. 2004; Sherrington et al. 2008, 2011)*. This indicates that improvements in muscle strength do not necessarily transfer to improved balance or reduced disability *(Skelton et al. 1995; Keysor & Jette 2001)*. In contrast, intervention programs which include balance training reveal a positive effect for the fall rate *(Rubenstein et al. 2000; Lord et al. 2003; Nitz & Choy 2004; Sherrington et al. 2008)* and have been reported to improve performance in clinical balance and mobility tests *(Wolfson et al. 1996; Steadman et al. 2003; Silsudapol et al. 2006; Madureira et al. 2007)*. However, balance training is not strictly defined and therefore not only the names for the training programs but also the contents of the balance training programs may vary. In general, training programs for the improvement of static and/or dynamic stability are designed to challenge the visual, vestibular and somatosensory systems and to manage different body positions. They include exercises in different stance positions (e.g. narrow/wide base of support, standing on variable surfaces) or during locomotion (e.g. tandem-walk, walking on instable surfaces) and some training programs further reduce sensory input. Regarding the outcome of the balance training programs, postural balance, measured by force platform measures, further improved after balance training regimens *(Wolfson et al. 1996; Rose & Clark 2000; Islam et al. 2004; for a review see Howe et al. 2011).* Additionally, older adults were able to increase their volitional maximum excursion of the limits of stability *(Islam et al. 2004).* This indicates an increased area of support which can be used to sustain balance during dynamic activities.

Regarding the neuronal explanatory approach of balance training, improved stance stability in consequence to a balance training program was correlated with decreased cortical excitability *(Taube et al. 2007)*. In response to balance training, the study by *Taube et al.* found a reduction in spinal excitability as well as in corticospinal and cortical excitability during the transcortically mediated long latency response *(Taube et al. 2007)*. The reduction in spinal excitability was thought to be caused by an increasing, supraspinal induced, pre-synaptic inhibition in consequence to balance training. Therefore it can be suggested that mainly supraspinal adaptations are responsible for the improved balance performance and that cortical plasticity plays an important role in the acquisition of balance skills *(Taube et al. 2007)*. The inhibition of spinal reflexes in balancing tasks was thought to be beneficial because of the reduction of reflex mediated joint oscillations and a shift of the movement control to higher centers *(Llewellyn et al. 1990; Solopova et al. 2003).* This shift may be used by the human system to compensate for enhanced afferent input and therefore to avoid exaggerated reflex responses *(Gruber et al. 2007)*. Inhibition of spinal reflexes seems to be enhanced when the postural task is more complex, indicating a higher need for supraspinal control *(Llewellyn et al. 1990; Earles et al. 2000)*. Furthermore, previous studies hypothesize that continuous and progressive training of motor tasks reduces the initial high activity of the motor cortex and enhances the contribution of subcortical regions like basal ganglia and cerebellum *(Floyer-Lea & Matthews 2004; Puttemans et al. 2005)*. With increasing automatization of the task by progressive training, neural control therefore shifts more and more to subcortical regions *(fig. 2-11)*.

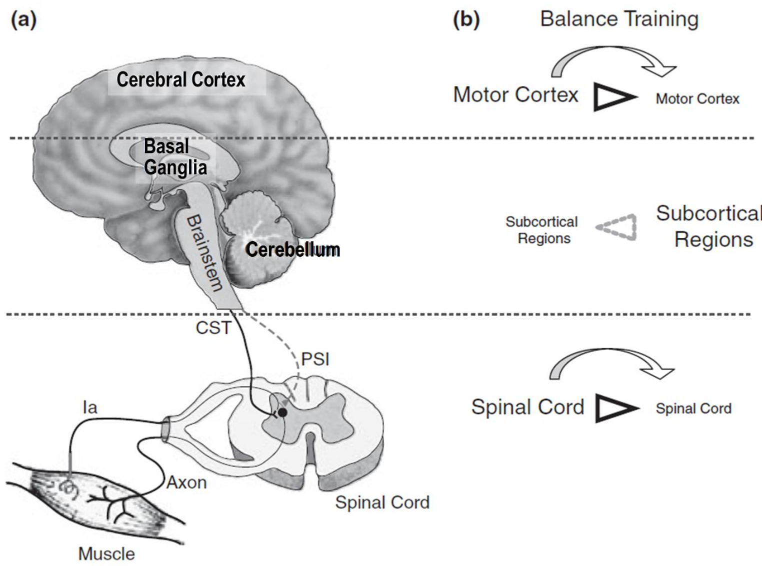

Figure 2-11: *Simplified illustration of spinal and supraspinal adaptations associated with balance training. a) Structures of the nervous system which are considered to be responsible for balance recovery. b) Balance training is thought to increase the supraspinal-induced presynaptic inhibition (PSI) and consequently to reduce the spinal reflex excitability. Cortical involvement is reduced with ongoing balance training whereas the involvement of subcortical regions increases more and more. Reprinted from Taube et al. (2008, p.111) with permission by Blackwell.*

With respect to the training adaptation on a neuronal level, elderly subjects show similar adaptations of the H-reflexes (down-regulation) after postural training compared to young adults *(Mynark & Koceja 2002)*. This reduction of the H-reflex amplitude is accompanied by a reduced postural sway area *(Mynark & Koceja 2002)*. However, reflex behavior of elderly was not modified when changing the position (supine vs. standing position) whereas young adults modulated the reflex output, showing differences in the ability to modulate the reflex system *(Koceja & Mynark 2000)*. In consequence to balance training yet, older adults also were able to modulate their reflex response. They demonstrated greater functional reflex responses in the prime movers and a decrease in the onset latencies as well as a decrease in the maximal angular velocity of the ankle joint complex while compensating perturbations *(Granacher et al. 2006)*. Those changes are ascribed to an improved conduction velocity of group-II-afferents and possibly also an enhanced muscle spindle sensitivity which together lead to an increased ankle joint stiffness during the perturbation *(Granacher et al. 2006)*. The enhanced reflex response here contrasts with the aforementioned studies which caused a reduction of spinal excitability. Neuronal adaptations to balance training therefore seem not always to be straightforward but rather context-dependent – the central nervous system learns to differentiate where reflex responses are reasonable: functional relevant reflex responses are increased whereas counterproductive responses are inhibited *(Taube 2012)*.

However, most intervention programs are based on exercises which focus on the control of balance during volitional movement. Yet, regarding fall risk in elderly, especially the motor performance in consequence to unpredictable perturbations is essential for the prevention of falls. Feed-forward mechanisms prior to the experience of a perturbation are important and serve to minimize the perturbation magnitude, but are not always possible or sufficient enough to eliminate the perturbation generated error. In this case, reactive feedback corrections have to restore the stability state. Volitional movements and perturbation-evoked reactions are controlled by different neural mechanisms *(Maki & McIlroy 1997)* – therefore, correspondent to the specificity-of-training principle, when aiming for better postural responses in consequence to perturbations, the training should include tasks which exercise the reactive behavior. According to that, several studies recently investigated the effect of perturbation-based balance training programs for older adults and included exercises which induce relative motion between the center of mass and the base of support to provoke recovery responses. However, several balance training programs already included at least some of those exercises such as standing on unsteady surface or handling of counter-rotations. Training regimens which focused on perturbation-based exercises revealed a significantly reduced frequency of multi-step reactions and foot collisions in response to surface translations compared to a control group *(Mansfield et al. 2010)*. Further, the time needed for step initiation and step completion as well as reaction time decreased *(Rogers et al. 2003; Jöbges et al. 2004; Shimada et al. 2004; Marigold et al. 2005)* whereas gait speed and stride length increased *(Jöbges et al. 2004)*.

Altogether, perturbation-based interventions rely on the *specificity-of-training principle* – the training therefore includes tasks which exercise the reactive behavior. Exercising specific tasks naturally is somewhat limited. Intervention programs are only able to include a small subset of daily activities and possible hazards, thus, it is not possible to exercise all potential perturbations in daily life.

Therefore, it is important to be able to transfer the learned behavior to different contexts. According to the training principles, possible transfer depends on the similarity between two tasks – this means that test performance is directly related to the similarity between characteristics of the practice and test condition *(Magill 2011)*. Since the common characteristic of the target movements is not a specific kinematic pattern but the reactive nature, exercise tasks in perturbation-based trainings focus on perturbed, reactive movements. More specifically, exercises in perturbation-based training are created to be specific to the particular demands of the task whereas the particular demand for the dynamic stability control is the control of the center of mass in relation to the base of support. This means that balance perturbations which challenge the control of this relationship are thought to be successful. The extent of *generalizability* depends on the characteristics of the performed tasks – learning new motor tasks seems to be more generalizable *(van Hedel et al. 2002)* than adapting to sensory-perceptual manipulations *(Earhart et al. 2002a,b)*. With respect to postural control, the generalization process requires a recalibration of motor control. According to *Bastian (2008)* and *Wang et al. (2011),* the central nervous system builds and updates its corresponding representation during this process to be able to predict the outcome by using feed-forward mechanisms and further to guide both, feed-forward adjustments and feedback responses *(Bastian 2008; Wang et al. 2011)*. (Neural) Representation here refers to the relation between motor commands and movements.

An internal representation evolves from motor practice and is used by the central nervous system to generalize motor adaptation to changing task objectives *(Shadmehr & Mussa-Ivaldi 1994; Wang et al. 2011)*. A more generalized internal representation is not dependent on specific effectors, tasks or environments and therefore motor transfer is more likely *(Morton et al. 2001; Wang et al. 2011)*. Transfer is thought to occur when the central nervous system recalibrates the non-specific and generalized representation of stability limits to the actual demands *(Wang et al. 2011)*. This means, that in consequence to specific training regimens, the central nervous system learns to adjust the motor commands and applied strategies according to the actual (dynamic) stability limits. The non-specific generalized representation of stability limits helps the central nervous system to assess the possible recovery strategies. When considering the generalization of recovery strategies, it is worthwhile to look for underlying common characteristics.

A closer look on the recovery strategies reveals different underlying mechanisms or strategies of the human system to control the center of mass in relation to the base of support. Exercising especially those mechanisms should exercise the control of postural stability. According to *Hof (2007)*, there are three distinct mechanisms which are applied for regaining dynamic stability, based on the equations of motion of the inverted pendulum model and on the equations of a multi-segment model *(Hof et al. 2005; Hof 2008)*. As aforementioned, the mechanisms of dynamic stability are *a) "increase of the base of support" or "moving the center of pressure", b) "counter-rotating segments around the center of mass" and c) "application of external force" (Hof 2007)*. Those three mechanisms are thought to be underlying strategies of the human system to achieve a stable posture, responsible for the control of dynamic stability. Thus, exercising these mechanisms should allow a generalization of the required strategies. The advantage of exercising mechanisms compared to exercising specific tasks is the flexible composition and adjustment of tasks in the training regimen. Furthermore, it is expected that exercising mechanisms should achieve not only a transfer effect to similar tasks, but a generalization effect which allows the application of the mechanisms of dynamic stability in various situations.

3. Purpose of the thesis

Summarizing, although some studies investigated the predictive and reactive adaptability in old adults, there is little information about the adaptability of older participants during walking. Especially the age-related reactive adaptation potential in dynamic stability control after perturbations during walking has not been investigated. Furthermore, it has not been investigated until now if and to what extent an exercise intervention of the mechanisms of dynamic stability may improve the stability state of old participants after unexpected perturbations during walking. Thus, the international literature reveals following open questions:

With regard to known, predictable environmental hazards, it is important to know if elderly adults are able to learn how to deal with those potential dangerous situations. Therefore, concerning the predictive adaptability of elderly to perturbations during walking, we want to know if the adaptation rate of young and older adults to perturbations during walking is similar or if the elderly need more time to show similar adaptations. In the course of several perturbations, we also want to know if elderly adults are able to show a similar adaptation magnitude compared to young individuals. Furthermore, until now it is not known if young and older adults apply similar postural strategies when adapting to predicted perturbations.

With reference to novel, unexpected perturbations, older adults have shown to have more problems to regain balance in comparison to young. Therefore, regarding possible intervention programs, it is essential to know if elderly are able to learn to resist balance after unpredictable perturbations. We do not know until now if old individuals are able to adapt with a similar adaptation rate and magnitude as young individuals in consequence to repeated perturbations during walking.

Further, not only the short-term retention of adaptation to experienced perturbations is relevant for prospective better handling of those perturbations, but also the long-term retention and the transfer effect to various contexts. If older adults are able to preserve a long-term knowledge about the handling of specific perturbations, training of those perturbations could help to better regain balance. Furthermore, we want to know if the training of underlying mechanisms for the control of dynamic stability may help to generalize the knowledge about the handling of particular perturbations.

More specifically, the included studies pursued following objectives:

1. The purpose of the *first study* was to investigate how old and young adults react in response to unexpected perturbations during walking and to compare the adaptive behavior of old and young adults to perturbations during walking. The hypothesis of this study states that older individuals show a reduced dynamic stability after an unexpected perturbation and a lower rate and magnitude of adjusted predictive motor behavior in comparison to young individuals.

2. The *second study* aimed to investigate the adaptability of young and older adults to repeated but unexpected perturbations during walking. On basis of previous studies we hypothesized that indeed both age-groups will show adaptive reactive behavior in consequence to repeated unexpected perturbations during walking. However, the old age group was suggested to show a slower reactive adaptation rate and lower reactive adaptation magnitude compared to the young.

3. Furthermore, the purpose of the *third study* was to investigate the effect of a training intervention on the recovery behavior of older adults after an unexpected perturbation during walking. The training intervention included exercises of the mechanisms of dynamic stability. We hypothesized that exercising the mechanisms of dynamic stability contributes to an improvement of the recovery performance after unexpected perturbations. Additionally, due to the effectiveness of multifactorial programs, we hypothesized that a combined intervention program which includes also strength training for the lower extremities would show even higher improvements in the recovery performance of old individuals.

The main characteristics of the three studies are listed in table 3-1.

Table 3-1: Main characteristics of the performed studies

Study	**Type of participants**	**Groups**	**Measurement of**	**chapter**
Bierbaum et al. 2010	Healthy, young & old	Young (n=10) & old (n=13)	Predictive dynamic stability	4
Bierbaum et al. 2011	Healthy, young & old	Young (n=14) & old (n=14)	Reactive dynamic stability	5
Bierbaum et al. 2012	Healthy, old (>65yrs.)	Stability training group (n=14), Mixed training group (n=14), Control group (n=12)	Reactive dynamic stability, strength	6

4. First Study: Adaptational responses in dynamic stability during disturbed walking in the elderly

Stefanie Bierbaum[1]
Andreas Peper[1]
Kiros Karamanidis[2]
Adamantios Arampatzis[1,3]

[1] Department of Training and Movement Sciences, Humboldt-University Berlin

[2] Institute of Biomechanics and Orthopaedics, German Sport University Cologne

[3] Center for Sports Science and Sports Medicine Berlin

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4.1 Abstract

The purpose of this study was to examine the age-related predictive and feedback adaptive locomotor improvements in the components of dynamic stability control during disturbed walking. Thirteen old (62-74yr) and ten young (23-30yr) male subjects performed a gait protocol on a gangway which included one covered element. By exchanging this element, the subjects walked either solely over hard surface or experienced a perturbation of the gait on the soft surface element. The gait protocol consisted of a baseline on hard surface and an adaptation phase with 19 trials on soft or hard (2^{nd}, 8^{th} and 19^{th}) surface. The investigation of dynamic stability was made by using the margin of stability (MS) which was calculated as the difference between base of support and the extrapolated center of mass (CM). Horizontal velocity of CM and its vertical projection in anterior-posterior direction as well as the eigenfrequency of an inverted pendulum generate the extrapolated CM. As a result of the first unexpected disturbance, MS was decreased in the step following the perturbation compared to baseline in both age-groups. This decrease was higher for the old participants compared to the young ones, indicating a more unstable position in the step after the perturbation for the elderly. In the following adaptation phase, MS returned to baseline in both age-groups. In the hard trial after the first unexpected perturbation, both age-groups increased MS at touchdown of the disturbed leg compared to baseline, reflecting fast predictive adjustments. Our findings show that the well-known age-related biological impairments did not inhibit the adaptive improvements in the components of dynamic stability in the elderly. However, the feedback corrections after the first unexpected perturbation were less effective for the elderly. This may increase the risk of falling.

4.2 Introduction

Epidemiological studies show that about 33% of the elderly people (>65 yrs.) experience at least one fall per year. Consequently fall related injuries in the ageing population represent a serious social and individual problem *(Blake et al. 1988; Tinetti et al. 1988)*. After unexpected perturbations, recovery performance seems to decrease notably for older individuals *(Wojcik et al. 1999; Pijnappels et al. 2005; Karamanidis & Arampatzis 2007),* which leads to a higher risk of falls within the elderly population. The decreased ability to recover of the elderly is attributed to an age-related reduction in muscle strength *(Schultz et. 1995; Grabiner et al. 2005; Karamanidis et al. 2008)*, modifications of the muscle activation timing *(Thelen et al. 1997)* as well as to lower muscular contraction velocities *(Larsson et al. 1979; Hortobágyi et al. 1995; Thelen et al. 1997)*. However, locomotor control can be modified by predictive and feedback based adaptive improvements of postural behavior. Humans have the ability to adapt to changes in the environmental conditions and thus to reduce differences between the preplanned motor outputs and the actual sensory input information through adjustments of the motor output. Changes in the environmental conditions require fast adjustments in motor control to achieve successful movement behavior.

Predictive feedforward adjustments are based on available knowledge about the intended movement *(for reaching movements see Thoroughman & Shadmehr 1999)* and it is believed that predictive locomotor responses depend mainly on supraspinal structures *(Earhart et al. 2002a; Morton & Bastian 2006)*. On the other hand, locomotor adjustments, which are based on feedback information, depend on knowledge received during the movement *(Thoroughman & Shadmehr 1999)*. These feedback adjustments in locomotor output after disturbances are believed to be mainly controlled by lower neural centers (i.e. spinal cord or brain stem, *Morton & Bastian 2006*).

Older adults show, compared to the young ones, a reduced capacity of the proprioceptive system as well as a reduced central processing capacity of afferent information *(Erni & Dietz 2001; Marigold & Patla 2002; Patel et al. 2009).* These age-related deficits may result in important consequences in the course of time of adaptive improvements or in the adaptation potential of the elderly population. Impairments of afferent sensory information may affect reactive as well as predictive adjustments during locomotion, because this information is important to adequately plan and execute the upcoming movements. The majority of falls in older adults occurs during gait after a sudden perturbation resulting from tripping or slipping *(Berg et al. 1997)*. However, there is little information in the literature about age-related adaptive responses after sudden perturbations while walking. After induced slips during sit-to-stand movements, *Pavol et al. (2004)* reported that although young participants adjust to a greater extent, the adaptational responses were similar between young and old adults. However, the after effects in the last study have been examined in the end of the adaptation phase (i.e. after 5 repeated slips) and therefore it is difficult to assess the rate of the predictive adaptational responses.

Based on the equations of motion of the inverted pendulum model *(Hof et al., 2005; Hof 2008)* as well as on the equations of a multi segment model *(Hof 2007)*, there are three mechanisms from a mechanical point of view which are responsible for regaining balance after a perturbation: a) by increasing the base of support (b) by counter-rotating segments around the center of mass (CM) and (c) by applying an external force, other than the ground reaction force (e.g. grasping). Considering those mechanisms, the effectiveness of dynamic stability control can be quantified *(Hof et al. 2005; Hof 2007, 2008)*. However, there is no information about age-related adaptive responses regarding the mechanisms of dynamic stability over the course of repeated perturbations while walking. Therefore, the purpose of the present study was to investigate how old and young people react after unexpected perturbations and how they adapt to repeated perturbations with regard to dynamic stability while walking. We hypothesized that older adults show impairments in dynamic stability after an unexpected perturbation and a decline in the rate and magnitude of predictive and feedback based adaptational responses compared to young adults.

4.3 Methods

4.3.1 Experimental design

Thirteen male elderly subjects (O: 67.4 ± 3.4 yrs.) and ten male young subjects (Y: 26.1 ± 2.9 yrs.) with similar body mass (O: 74 ± 7 kg; Y: 70 ± 6 kg) and body height (O: 174 ± 5 cm; Y: 176 ± 7 cm) participated in the study after giving informed consent to the experimental procedure according to the rules of the local scientific board. All subjects were physically active and without neurological or musculoskeletal impairments.

Walk-to-run transition velocity (WRV) was determined for each participant on the basis of four to six walking trials before the main experiments. Subjects first had to walk at their self-chosen walking speed. After that velocity was increased during the first accommodation trials to evoke the maximal gait velocity. WRV was defined as the maximal gait velocity with double stance phase, checked through kinetic measurements. Velocity was monitored by light barriers enclosing the capture volume and via kinematic data recorded by the Vicon system. Young participants achieved a higher maximal gait velocity than the old ones ($young_{mean}$: 3.43 ± 0.27 m/s, range: 2.90 - 3.74 m/s; old_{mean}: 2.95 ± 0.3 m/s, range: 2.56 - 3.4 m/s). Once the maximal gait velocity was determined, every subject had to walk at 60% of his WRV throughout all following trials. This was done to obtain similar demands for the neuromuscular system of the two age-groups during the experiments *(Neptune & Sasaki 2005)*. The target velocity (60% of WRV) was controlled by a stick, which moved in front of the subjects along the gangway (15 m) and by the light barriers.

An exchangeable element (90 x 60 x 20 cm^3) was placed halfway in the gangway. This element was hidden by a cover sheet to be able to change the surface from hard to soft and vice versa without the knowledge of the subjects. The upper surface of the soft mat consisted of relatively hard rubber material (0.8 cm). Beneath that, the soft element (17 cm depth) was made of foam. The force deformation characteristic of the soft surface was non-linear. In average the deformation of the surface within the experiment was about 10 cm for the old and young participants. Velocity and step length were kept quite similar throughout the experimental trials so that the participants would always step with their right leg on the exchangeable element.

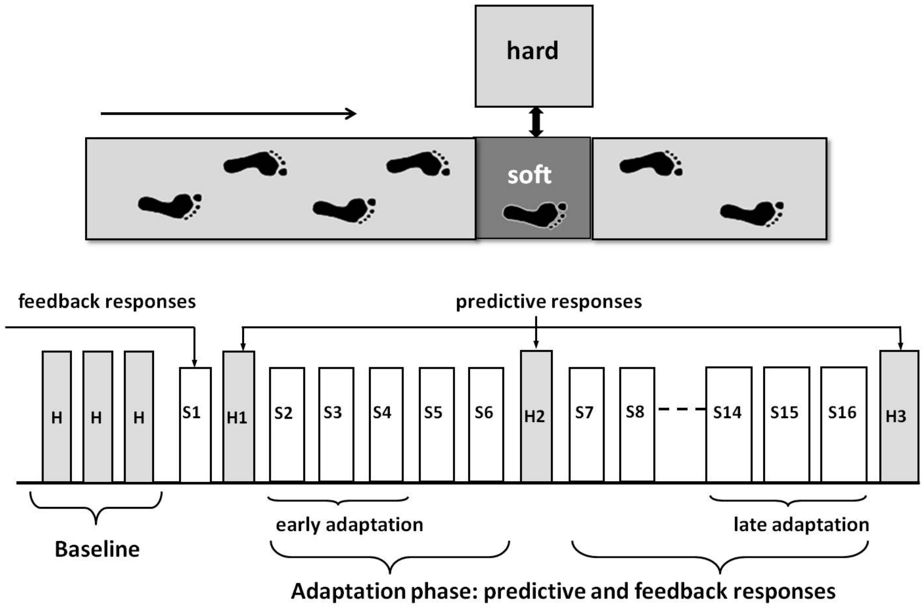

Figure 4-1: *Experimental protocol of hard and soft surface trials. The gangway included one covered, exchangeable element which allowed changing the surface condition from hard to soft and vice versa without the knowledge of the participants. Baseline, which contained 3 trials on hard surface (H), was followed by one unexpected soft surface trial (S1). The next hard surface trial (H1) and the hard surface trials after 5 (H2) and 10 (H3) soft trials respectively were used to analyze predictive responses (after effects). The soft surface trials documented the adaptation phase. The subjects started the baseline trials with the information that they would have to expect hard surface and continued after the first soft trial with the new information that there would stay the soft surface.*

The gait protocol consisted of 22 gait trials starting with three baseline trials on the hard surface (baseline) and was followed by 19 gait trials in the adaptation phase *(fig. 4-1)*. Before starting the experiment, the participants got the information that the surface could change without further warnings. After the first soft surface trial they just were told that the surface will stay for all following trials “soft”. No further advices were given. The first trial of the adaptation phase included an unexpected perturbation by the soft surface. This trial was used to detect feedback responses, because the subjects could not anticipate the first unexpected perturbation, but were only able to react after the perturbation to regain balance. The three hard trials within the adaptation phase were performed to examine after effects and thus predictive responses while walking. After effects are criterions for predictive motor adaptation *(Martin et al. 1996b; Fernandez-Ruiz & Diaz 1999)*. These three hard trials have been performed to assess the rate and magnitude of the predictive adaptational responses within the experiment.

4.3.2 Quantification of dynamic stability control

Ground reaction forces of the step before and on the exchangeable gangway element were collected at a sampling rate of 1080 Hz with two Kistler force plates (60 x 90 cm, Kistler, Winterthur, Switzerland). Disturbed leg refers to the leg, that stepped on the exchangeable element and the recovery leg is defined as the leg that helped to regain balance after the perturbation. Kinematic data were recorded with the Vicon motion capture system (Model 624, Vicon, Oxford, UK) using 12 cameras operating at 120 Hz. Thirty-eight reflective markers (diameter 14 mm) were used to track the whole body kinematics. The marker trajectories were smoothed, using a Woltring filter routine *(Woltring 1986)* with a noise level of 10 mm^2. Segmental masses and the location of the segment centers of mass were calculated through the Plug-In-Gait model, based on the data reported by *Dempster et al. (1959)*.

To recognize predictive as well as feedback responses in dynamic stability control we used the "extrapolated center of mass" concept formulated by *Hof et al. (2005) (fig.4-2)*. The margin of stability as a criterion for the state of stability of the human body was calculated according to Hof et al. (2005):

$$b_x = U_{\max} - X_{CM}$$

where b_x indicates the margin of stability in the anterior-posterior direction, U_{max} is the anterior boundary of the base of support and X_{CM} is the position of the extrapolated center of mass in the anterior-posterior direction $(X_{CM} = P_{XCM} + \frac{V_{XCM}}{\sqrt{g/l}})$. P_{XCM} is the horizontal (anteroposterior) component of the projection of the center of mass (CM) to the ground, V_{XCM} is the horizontal CM velocity and the term $\sqrt{g/l}$ presents the eigenfrequency of a system of length l (inverted pendulum model). Margin of stability was determined in the anteroposterior direction since the extrapolated center of mass mainly shifts in the anterior direction after the perturbation due to the increased velocity of center of mass. Margin of stability and the components of the dynamic stability (base of support, horizontal component of the projected CM to the ground, horizontal velocity of the CM and term $\sqrt{g/l}$) were used to quantify the dynamic stability while walking. Their validity was demonstrated in previous studies *(Mademli et al. 2007; Arampatzis et al. 2008; Karamanidis et al. 2008)*. For each trial three time points were identified: a) touchdown of the disturbed leg (before experiencing the perturbation; TDdist), b) touchdown of the recovery leg (TDrec) and c) the first local maximum of the anteroposterior center of mass velocity after the perturbation *(fig. 4-3)*. The third time point was chosen to assess the direct consequences of the disturbance.

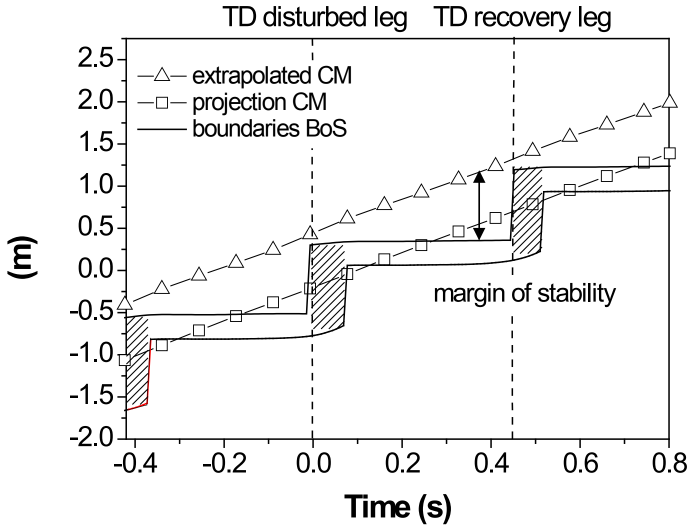

Figure 4-2: *Parameters of dynamic stability control using the example of one baseline trial: extrapolated center of mass (CM), projection of CM to the ground and boundaries of the base of support (BoS) in the anterior-posterior direction. The top solid line shows the anterior boundary and the bottom line the posterior boundary of the BoS. The shaded area indicates the double stance phase.*
Base of support refers to the distance from the anterior to the posterior boundary of the base of support for both feet whereas the positions of the foot markers at the calcanei and the 2nd metatarsal bone were registered relative to individual sketches of the subjects´ feet so that the boundaries of the base of support could be calculated out of the kinematic data. Positive values of the margin of stability represent a stable position of the human body, negative values indicate an unstable position, i.e. the human system has to perform additional actions to maintain balance. Note that in this specific walking velocity (~1.8 m/s) the extrapolated CM is located permanently outside of the base of support (i.e. negative values of the margin of stability).

Differences in margin of stability of the disturbed leg in the hard trials compared to the baseline condition quantify the predictive adjustments of the participants during walking. Comparisons of the margin of stability values at TDrec between the first unexpected soft trial and the baseline trials quantify the quality of the feedback responses, whereas the comparisons of the following soft trials to the baseline demonstrate the success of feedback as well as predictive responses.

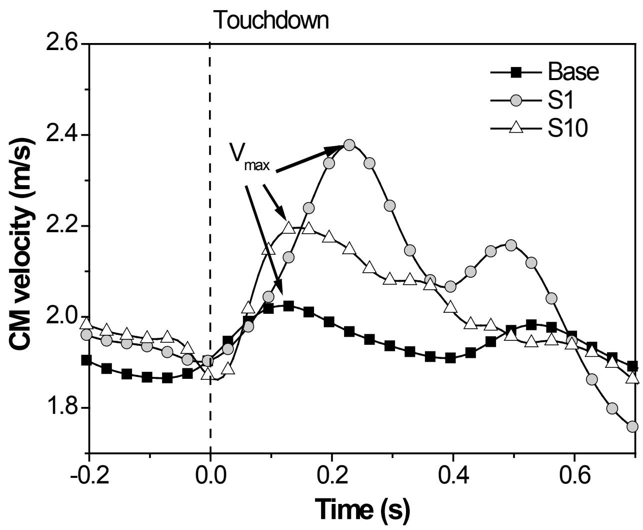

Figure 4-3: *Horizontal velocity of center of mass at a baseline trial, the first unexpected soft (soft1) and a further soft trial (soft10). The first local maximum in the horizontal center of mass velocity after touchdown of the disturbed leg was used as an indication for the immediate consequences of the disturbance.*

4.3.3 Statistics

The mean values of the three baseline trials on hard surface were used to establish the base level. The soft trials two to four as well as the soft trials 14 to 16 of the adaptation phase were pooled together as representatives of the early and late adaptation phase respectively. Those mean values of the baseline and early and late adaptation phase as well as the first trial on the soft surface and the following hard trials during the adaptation phase were included in the analysis. This was done to be able to analyze the adaptational responses to the soft surface (S1, early and late adaptation phase) and, even more important in our setting, to investigate predictive adaptational behavior in anticipation of the soft surface in conditions which are similar to the baseline (hard trials in-between).

A two-way analysis of variance with the fixed variables age-group and repetition and the gait velocity of the participants as covariate was used to examine the age (young, old) and repetition related differences in the analyzed dynamic stability parameters (margin of stability, base of support, position of the extrapolated CM, horizontal component of the projection of the CM to the ground, horizontal CM-velocity and the term $\sqrt{g/l}$). Anthropometric data was compared with a *t*-test for non-dependent samples. The level of significance for all statistical comparisons was set to $\alpha = 0.05$.

4.4 Results

4.4.1 Touchdown of disturbed leg

The margin of stability at touchdown of the disturbed leg (TDdist) increased in both groups (no significant age effect, p = 0.81) for the hard trial (H1) directly after the first unexpected perturbation (first soft trial) compared to the baseline *(fig. 4-4)*. This means that both age-groups showed after effects very quickly and thus predictive adaptational responses following the first experience of the perturbation. At touchdown of the first hard trial, the horizontal projection of the CM to the ground decreased in both age-groups compared to the baseline (more posterior position of the CM projection relative to the anterior boundary of base of support), while the base of support, the horizontal velocity of CM and the term $\sqrt{g/l}$ remained unchanged *(table 4-1)*. The position of the extrapolated CM in the first hard trial showed a tendency (p = 0.07) to lower values compared to the baseline. In the other two hard trials during the adaptation phase the participants of both age-groups showed an increase in the base of support compared to the baseline and no differences in the other components of dynamic stability *(table 4-1)*. The age-related differences in the values originate from the different gait velocity for the two age-groups.

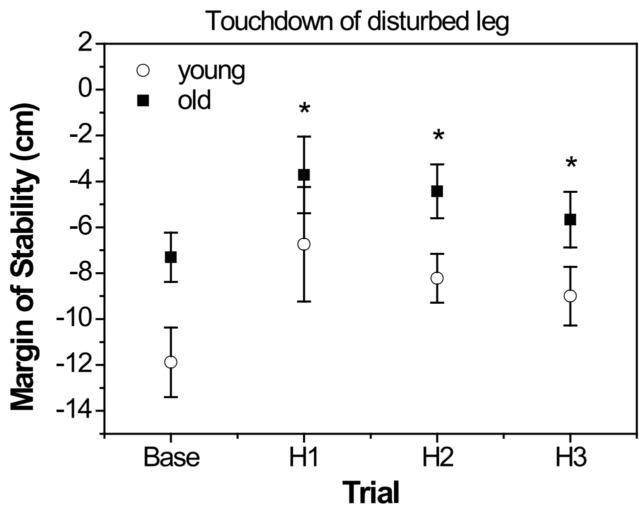

Figure 4-4: *Mean values and standard error of mean of Margin of stability at touchdown of the disturbed leg in the baseline and the hard trials (H1-H3) during the adaptation phase. Note that the different values for the margin of stability for the young and old participants originated from the different walking velocity. *: statistically. sign. difference to the baseline (p < 0.05).*

Compared to the baseline, the margin of stability as well as the base of support at TDdist showed higher values in the early and late adaptation phases in both age groups *(fig. 4-5)*. This behavior indicates that the reason for the increase in the margin of stability in the early and late adaptation phase was the increase in the base of support. The first local maximum of horizontal CM velocity after touchdown of the disturbed leg showed the highest values after the first unexpected perturbation *(fig. 4-6)*. After the first unexpected soft surface trial, the first local maximum of horizontal CM velocity decreased significantly ($p< 0.05$), but remained on a higher level compared to baseline values. The braking horizontal ground reaction forces of the disturbed leg decreased dramatically in the first unexpected perturbation compared to the base line in both age groups *(fig. 4-7)*. In the early and late adaptation phases the horizontal braking forces showed an increase compared to the first unexpected soft trial, but remained lower compared to the baseline *(fig. 4-7)*.

Table 4-1: *Means ± SD of the extrapolated center of mass (CM), horizontal component of the projected CM to the ground, horizontal velocity of the CM and term* $\sqrt{(g/l)}$ *(g: acceleration of gravity, l: distance between CM and center of ankle joint in sagittal plane) for the young and elderly people at baseline and the following experimental hard trials. Extrapolated CM and projection of CM are calculated in reference to the anterior boundary of the hindlimb at touchdown of the disturbed leg.*

	Young				**Old**			
	Baseline	**Hard1**	**Hard2**	**Hard3**	**Baseline**	**Hard1**	**Hard2**	**Hard3**
Base of support [cm]	87.8 ± 5.0	89.6 ± 5.0	92.1 ± 5.1 *	92.0 ± 5.6 *	82.9 ± 5.6	82.4 ± 7.7	84.1 ± 5.9 *	84.0 ± 6.4 *
Extrapol. CM [cm]	99.7 ± 7.7	96.3 ± 10.5	100.4 ± 5.8	101.0 ± 7.6	90.2 ± 8.1	86.1 ± 11.2	88.2 ± 9.1	89.7 ± 7.0
Proj. CM [cm]	36.2 ± 3.1	33.2 ± 7.8 *	37.4 ± 2.9	37.8 ± 3.8	33.72 ± 4.0	30.9 ± 5.88 *	31.3 ± 4.5	32.3 ± 4.0
Horizont. V_{CM} [m/s]	2.01 ± 0.13	2.00 ± 0.11	1.99 ± 0.13	1.99 ± 0.14	1.79 ± 0.14	1.75 ± 0.16	1.82 ± 0.15	1.82 ± 0.12
Term [s^{-1}]	3.19 ± 0.1	3.17 ± 0.09	3.17 ± 0.08	3.16 ± 0.1	3.18 ± 0.07	3.18 ± 0.08	3.2 ± 0.09	3.17 ± 0.06

** Statistically significant difference to the baseline values ($p < 0.05$).*

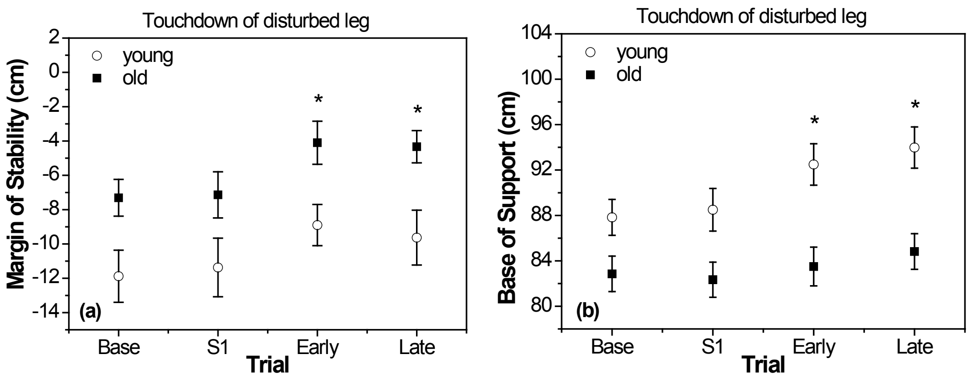

Figure 4-5: *Mean and standard error of mean of margin of stability (a) and base of support (b) at touchdown of the disturbed leg. The parameters of dynamic stability control are shown at baseline, the first unexpected soft surface (S1) and at the early (S2-S4) and late (S14-16) adaptation phase. *: statistically significant differences to the baseline values (p < 0.05).*

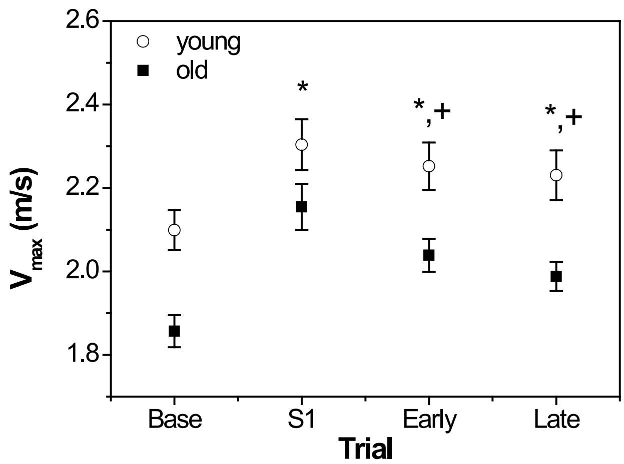

Figure 4-6: *Mean values and standard error of mean of the first local maximum in the horizontal center of mass velocity (V_{max}) after the perturbation. The figure shows the mean of baseline trials on hard surface (Base), the first experimental trial on unexpected soft surface (S1) and the early and late adaptation phase on soft surface. *: statistically significant difference to the baseline values (p < 0.05), +: statistically significant difference to S1 (p<0.05).*

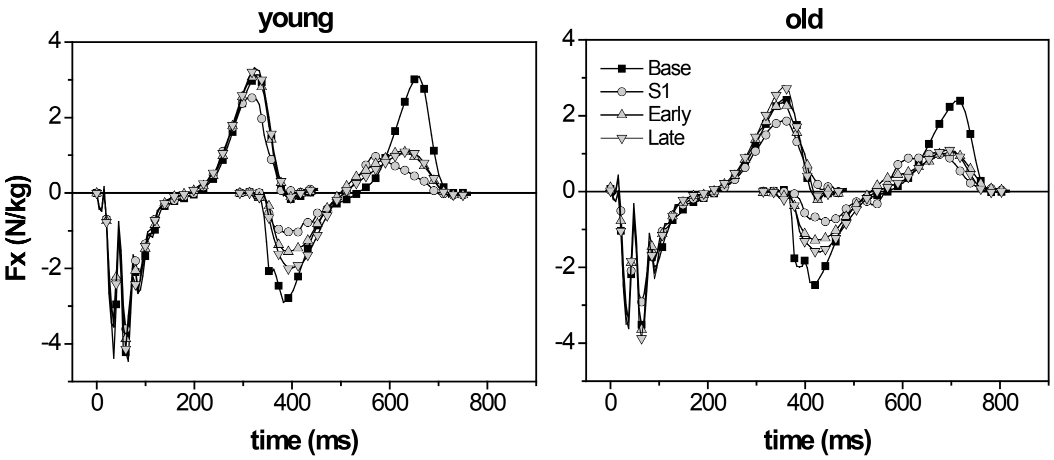

Figure 4-7: *Mean values of the horizontal forces of forceplate 1 (step before perturbation) and forceplate 2 (step on the exchangeable (soft/hard) element). Presented are baseline (Base), the first unexpected soft surface (S1), early (Early, S2-4) and late (Late, S14-16) adaptation phases.*

4.4.2 Touchdown of recovery leg

In the step following the perturbation *(touchdown of recovery leg, TDrec)* margin of stability was decreased for the first unexpected perturbation compared to the baseline values in both age-groups *(fig. 4-8)*. However, the decrease in the margin of stability at TDrec was higher for the elderly participants compared to the young ones, indicating a more unstable position at touchdown for the elderly *(fig. 4-8)*. Both age-groups increased significantly their base of support at TDrec compared to the baseline in order to regain balance during the first soft trial ($p < 0.05$, table 4-2). This increase in the base of support was less pronounced for the elderly subjects and caused the more unstable position of this group.

In the early and late adaptation phases on the soft surface, margin of stability at TDrec increased for both age-groups compared to the values of the first unexpected perturbation, recovering to values of the baseline in the late adaptation phase *(fig. 4-8)*. Base of support, extrapolated CM and the projected CM at TDrec showed increased values in the early and late adaptation phase compared to the baseline *(table 4-2)*. The horizontal CM velocity returned to the baseline level in the late adaptation phase.

Table 4-2: *Means ± SD of the extrapolated center of mass (CM), horizontal component of the projected CM to the ground, horizontal velocity of the CM and term $\sqrt{(g/l)}$ (g: acceleration of gravity, l: distance between CM and center of ankle joint in sagittal plane) for the young and elderly people at baseline and the experimental soft trials. Base of support, extrapolated CM and projection of CM are calculated in reference to the anterior boundary of the hindlimb at touchdown of the disturbed leg.*

	Young				**Old**			
	Baseline	**Soft1**	**Early**	**Late**	**Baseline**	**Soft1**	**Early**	**Late**
Base of support [cm]	90.1 ± 5.1	104.8 ± 9.2*,°	103.6 ± 8.1*	101.2 ± 6.5*	82.7 ± 6.3	87.2 ± 12.2*,°	89.8 ± 9.7*	92.6 ± 6.3*
Extrapol. CM [cm]	102.9 ± 7.9	121.1 ± 14.9*	118.6 ± 10.0*	116.3 ± 11.0*	91.2 ± 8.6	106.8 ± 16.8*	100.2 ± 12.1*	100.1 ± 7.9*
Proj. CM [cm]	39.7 ± 3.7	53.8 ± 8.4*	51.5 ± 5.2*	51.3 ± 5.6*	34.7 ± 3.8	45.3 ± 10.2*	42.1 ± 8.7*	42.6 ± 6.2*
Horizont. V_{CM} [m/s]	2.01 ± 0.12	2.1 ± 0.2*	2.11 ± 0.16*	2.04 ± 0.17	1.80 ± 0.15	1.93 ± 0.24*	1.84 ± 0.12*	1.81 ± 0.12
Term [s^{-1}]	3.18 ± 0.98	3.14 ± 0.17	3.14 ± 0.1	3.14 ± 0.11	3.19 ± 0.07	3.14 ± 0.08	3.17 ± 0.09	3.16 ± 0.07

* Statistically significant difference to the baseline values ($p < 0.05$).
° Statistically significant age interaction ($p < 0.05$).

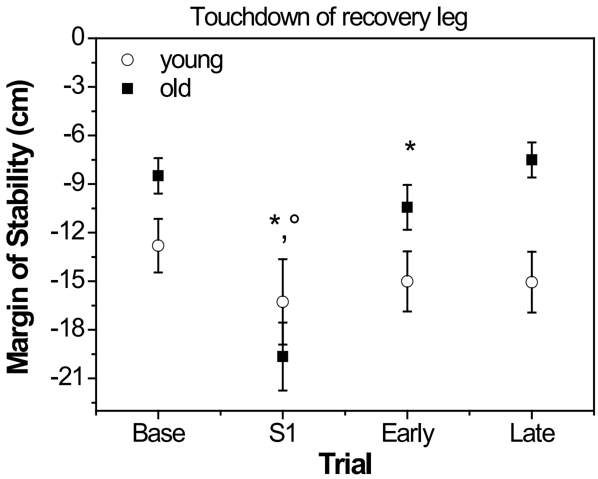

Figure 4-8: *Mean and standard error of mean of margin of stability at touchdown of the recovery leg after the perturbation. *: statistically sign. difference to the baseline values; °: statistically sign. age interaction (p < 0.05).*

4.5 Discussion

We investigated the questions of how old and young male participants react to unexpected environmental perturbations during gait with respect to dynamic stability control, how they adapt after repeated exposure to these environmental perturbations and how quickly feedback information is transferred into feedforward (predictive) adjustments. We hypothesized that old adults will show (a) impaired dynamic stability control after an unexpected perturbation, and (b) a decline in the rate and magnitude of predictive and feedback adaptational responses compared to young ones.

During the first unexpected perturbation, both age-groups showed an increase in the base of support at the step after the perturbation which served as a feedback response to correct the more unstable body position compared to the habitual body-stability at baseline. These feedback corrections were more effective for the younger adults which was reflected in the higher increase in the base of support compared to the elderly, leading to a lower decrease in the margin of stability at touchdown after the perturbation (TDrec). After a sudden perturbation a reasonable challenge of the central nervous system is to execute successful postural corrections by selecting motor plans which include mechanisms responsible for maintaining the dynamic stability. The feedback postural corrections at TD of the recovery limb were less effective for the old participants compared to the young ones after the unexpected soft trial. The effectiveness of the feedback-based corrections can be affected by the magnitude of the disturbance. Direct after the perturbation a higher disturbance increases the challenge for postural feedback corrections. Assuming a greater disturbance after the unexpected perturbation for the old adults compared to the young ones, the less effective reactive responses might be originated in heterogeneous acute disturbance levels. Therefore, we examined the changes in the first local maximum of the horizontal CM-velocity after the perturbation as a criterion for the magnitude of the acute consequences of the perturbation. However, our results confirm that the acute consequences in the dynamic stability after the disturbance were not different between the two age-groups. The first local maximum of the horizontal CM-velocity after the perturbation did not show any age effects indicating similar changes in dynamic stability in both groups. Even if reactive behavior should have been generated until the first local maximum of the horizontal CM-velocity, the effect on the magnitude of this value was similar for both age-groups. Therefore the age related differences of the feedback corrections after the first unexpected perturbation did not result from heterogeneous disturbance level.

Human motor behavior is the consequence of a combined effect between the capacities of the central nervous and the musculoskeletal systems *(Scott 2004)*. It is well known that old adults compared to the young ones show a reduction in the force generating capacity of the muscles *(Criswell et al. 1997; Frontera et al. 2000)* as well as a reduction in the central processing capacity of afferent information *(Light 1990; Prince et al. 1997).* There is evidence that the cost of postural adjustments is higher in the old compared to the young adults due to the need of more comprehensive cognitive processing in the elderly *(Teasdale et al. 1993)*. Additionally it is a well-known phenomenon that the contraction velocity of skeletal muscles is reduced with age *(D'Antona et al., 2003)*, which can alter the capacity to produce rapid balance corrections in the elderly population *(Thelen et al., 1997; Karamanidis & Arampatzis, 2007)*. Our data show that the lower stability of the old adults compared to the young ones at touchdown of the recovery leg is related to a lesser increase in the base of support for the elderly. Therefore, it is reasonable to suggest that the age-related deficits in the neuromuscular system are the reason for the less effective feedback corrections in the old participants while walking. However, in this study it is not possible to quantify the contribution of the muscular and central processing capacities on the measured age-related differences in stability control.

In the first unexpected perturbation the corrections in dynamic stability were inherently feedback adjustments because of the missing information about the modality of the disturbance beforehand. In the next trial (H1) both groups made use of the knowledge from the first perturbation, resulting in an increase in the margin of stability (i.e. a more stable position) at touchdown of the disturbed leg in a feedforward manner. The same after effects (i.e. increase in the margin of stability) have been found in the following two hard trials (H2 and H3). However, in the first hard trial the increase in margin of stability at touchdown of the disturbed leg occurred rather by a more backward position of the CM compared to the baseline. Though, the increase in margin of stability in the trials H2 and H3 trials happened by an increase in BS before the perturbation. This different locomotor strategy suggests that the adaptation in the first hard trial is not complete and that the participants performed the walking trial more carefully. The consequences and the feedback responses after the first unexpected perturbation induced the increase of the margin of stability in a predictive feedforward manner. The decrease of the first maximum of the CM-velocity after the disturbance in the early and late adaptation phase shows that the examined young and old adults adapted predictive adjustments in a similar manner. The reason for the decrease in the horizontal CM-velocity during the adaptation phase was the increase of the braking forces of the disturbed limb compared to the first unexpected trial. The increase of the horizontal braking forces of the disturbed limb resulted by the higher base of support (i.e. greater step) before the disturbance and thus because of the predictive responses. Compared to the first unexpected perturbation, base of support at touchdown of the recovery leg in the early and late adaptation phase remains invariant. Nevertheless the margin of stability at touchdown of the recovery leg increased in the early adaptation phase compared to the first unexpected perturbation and reached baseline values in the late adaptation phase. These findings show that the predictive adjustments improve the effectiveness of the reactive feedback responses.

The predictive improvements in margin of stability while walking happened very quickly (already the first hard trial showed after effects) and without any age-related differences. The adaptational improvements of the delayed motor responses (i.e. after the perturbation) to correct the state of stability were also similar in both old and young adults. Therefore, we can argue that the sensorimotor adaptation potential of dynamic stability control related to perturbations during gait is not reduced in the elderly population. The major component of the adaptation process was a modification of the boundaries of the base of support compared to the baseline (i.e. increase of base of support before and after the perturbation).

In conclusion, our findings demonstrated (a) that older adults showed deficits in the components of dynamic stability after an unexpected perturbation of similar demand compared to the young adults while walking and (b) that the age-related biological impairments of the human system do not inhibit the formation of adaptive improvements in the dynamic stability control after repetitive perturbations while walking. The above results may have important impacts on the quality of life in the elderly population. The reduced ability to control the dynamic stability after an unexpected perturbation of the old compared to the young adults may increase the risk of falling in the elderly during daily activities. However, the old adults achieved similar improvements in the adaptation rate and magnitude compared to the young ones during the disturbed walking protocol. This means that the adaptation potential related to perturbations does not decrease due to age. Therefore practicing tasks including the mechanisms responsible for dynamic stability control may allow older adults to use these mechanisms during sudden perturbations or tripping, improving their ability to recover without falling.

Conflict of interest statement

The authors disclose any financial and personal relationships with other people or organisations that could inappropriately influence (bias) their work.

4.6 References

Arampatzis, A., Karamanidis, K., and Mademli, L. (2008). Deficits in the way to achieve balance related to mechanisms of dynamic stability control in the elderly. J Biomech *41*, 1754–1761.

Berg, W.P., Alessio, H.M., Mills, E.M., and Tong, C. (1997). Circumstances and consequences of falls in independent community-dwelling older adults. Age Ageing *26*, 261–268.

Blake, A.J., Morgan, K., Bendall, M.J., Dallosso, H., Ebrahim, S.B., Arie, T.H., Fentem, P.H., and Bassey, E.J. (1988). Falls by elderly people at home: prevalence and associated factors. Age Ageing *17*, 365–372.

Criswell, D.S., Powers, S.K., Herb, R.A., and Dodd, S.L. (1997). Mechanism of specific force deficit in the senescent rat diaphragm. Respir Physiol *107*, 149–155.

D'Antona, G., Pellegrino, M.A., Adami, R., Rossi, R., Carlizzi, C.N., Canepari, M., Saltin, B., and Bottinelli, R. (2003). The effect of ageing and immobilization on structure and function of human skeletal muscle fibres. J Physiol (Lond.) *552*, 499–511.

Dempster, W.T., Gabel, W.C., and Felts, W.J. (1959). The anthropometry of the manual work space for the seated subject. Am. J Phys Anthropol *17*, 289–317.

Earhart, G.M., Fletcher, W.A., Horak, F.B., Block, E.W., Weber, K.D., Suchowersky, O., and Melvill Jones, G. (2002a). Does the cerebellum play a role in podokinetic adaptation? Exp Brain Res *146*, 538–542.

Erni, T., and Dietz, V. (2001). Obstacle avoidance during human walking: learning rate and cross-modal transfer. J Physiol (Lond.) *534*, 303–312.

Fernández-Ruiz, J., and Díaz, R. (1999). Prism adaptation and aftereffect: specifying the properties of a procedural memory system. Learn Mem *6*, 47–53.

Frontera, W.R., Suh, D., Krivickas, L.S., Hughes, V.A., Goldstein, R., and Roubenoff, R. (2000). Skeletal muscle fiber quality in older men and women. Am J Physiol, Cell Physiol *279*, C611–618.

Grabiner, M.D., Owings, T.M., and Pavol, M.J. (2005). Lower extremity strength plays only a small role in determining the maximum recoverable lean angle in older adults. J Gerontol A Biol Sci Med Sci *60*, 1447–1450.

Hof, A.L. (2007). The equations of motion for a standing human reveal three mechanisms for balance. J Biomech *40*, 451–457.

Hof, A.L. (2008). The "extrapolated center of mass" concept suggests a simple control of balance in walking. Hum Mov Sci *27*, 112–125.

Hof, A.L., Gazendam, M.G.J., and Sinke, W.E. (2005). The condition for dynamic stability. J Biomech *38*, 1–8.

Hortobágyi, T., Zheng, D., Weidner, M., Lambert, N.J., Westbrook, S., and Houmard, J.A. (1995). The influence of aging on muscle strength and muscle fiber characteristics with special reference to eccentric strength. J Gerontol A Biol Sci Med Sci *50*, B399–406.

Karamanidis, K., and Arampatzis, A. (2007). Age-related degeneration in leg-extensor muscle-tendon units decreases recovery performance after a forward fall: compensation with running experience. Eur J Appl Physiol *99*, 73–85.

Karamanidis, K., Arampatzis, A., and Mademli, L. (2008). Age-related deficit in dynamic stability control after forward falls is affected by muscle strength and tendon stiffness. J Electromyogr Kinesiol *18*, 980–989.

Larsson, L., Grimby, G., and Karlsson, J. (1979). Muscle strength and speed of movement in relation to age and muscle morphology. J Appl Physiol *46*, 451 –456.

Light, K.E. (1990). Information processing for motor performance in aging adults. Phys Ther *70*, 820–826.

Mademli, L., Arampatzis, A., and Karamanidis, K. (2008). Dynamic stability control in forward falls: postural corrections after muscle fatigue in young and older adults. Eur J Appl Physiol *103*, 295–306.

Marigold, D.S., and Patla, A.E. (2002). Strategies for dynamic stability during locomotion on a slippery surface: effects of prior experience and knowledge. J Neurophysiol *88*, 339–353.

Martin, T.A., Keating, J.G., Goodkin, H.P., Bastian, A.J., and Thach, W.T. (1996b). Throwing while looking through prisms. II. Specificity and storage of multiple gaze-throw calibrations. Brain *119 (Pt 4)*, 1199–1211.

Morton, S.M., and Bastian, A.J. (2006). Cerebellar contributions to locomotor adaptations during splitbelt treadmill walking. J Neurosci *26*, 9107–9116.

Neptune, R.R., and Sasaki, K. (2005). Ankle plantar flexor force production is an important determinant of the preferred walk-to-run transition speed. J Exp Biol *208*, 799–808.

Patel, M., Magnusson, M., Kristinsdottir, E., and Fransson, P.-A. (2009). The contribution of mechanoreceptive sensation on stability and adaptation in the young and elderly. Eur J Appl Physiol *105*, 167–173.

Pavol, M.J., Runtz, E.F., and Pai, Y.-C. (2004). Young and older adults exhibit proactive and reactive adaptations to repeated slip exposure. J Gerontol A Biol Sci Med Sci *59*, 494–502.

Pijnappels, M., Bobbert, M.F., and Dieën, J.H. van (2005). Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. Gait & Posture *21*, 388–394.

Prince, F., Corriveau, H., Hébert, R., and Winter, D.A. (1997). Gait in the elderly. Gait & Posture *5*, 128–135.

Schultz, A.B. (1995). Muscle function and mobility biomechanics in the elderly: an overview of some recent research. J Gerontol A Biol Sci Med Sci *50 Spec No*, 60–63.

Scott, S.H. (2004). Optimal feedback control and the neural basis of volitional motor control. Nat Rev Neurosci *5*, 532–546.

Teasdale, N., Bard, C., LaRue, J., and Fleury, M. (1993). On the cognitive penetrability of posture control. Exp Aging Res *19*, 1–13.

Thelen, D.G., Wojcik, L.A., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (1997). Age Differences in Using a Rapid Step To Regain Balance During a Forward Fall. J Gerontol A Biol Sci Med Sci *52A*, M8–M13.

Thoroughman, K.A., and Shadmehr, R. (1999). Electromyographic correlates of learning an internal model of reaching movements. J Neurosci *19*, 8573–8588.

Tinetti, M.E., Speechley, M., and Ginter, S.F. (1988). Risk factors for falls among elderly persons living in the community. N Engl J Med *319*, 1701–1707.

Wojcik, L.A., Thelen, D.G., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (2001). Age and gender differences in peak lower extremity joint torques and ranges of motion used during single-step balance recovery from a forward fall. J Biomech *34*, 67–73.

Woltring, H.J. (1986). A Fortran package for generalized, cross-validatory spline smoothing and differentiation. Adv Eng Software (1978) *8*, 104–113.

5. Second study: Adaptive feedback potential in dynamic stability during disturbed walking in the elderly

Stefanie Bierbaum[1]
Andreas Peper[1]
Kiros Karamanidis[2]
Adamantios Arampatzis[1,3]

[1] Department of Training and Movement Sciences, Humboldt-University Berlin

[2] Institute of Biomechanics and Orthopaedics, German Sport University Cologne

[3] Center for Sports Science and Sports Medicine Berlin (CSSB)

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5.1 Abstract

After perturbation of the gait, feedback information may help regaining balance adequately, but it remains unknown whether adaptive feedback responses are possible after repetitive and unexpected perturbations during gait and if there are age-related differences. Prior experience may contribute to improved reactive behavior.

Fourteen old (59-73yrs) and fourteen young (22-31yrs) males walked on a walkway which included one covered element. By exchanging this element participants either stepped on hard surface or unexpectedly on soft surface which caused a perturbation in gait. The gait protocol contained 5 unexpected soft trials to quantify the reactive adaptation. Each soft trial was followed by 4-8 hard trials to generate a “wash-out” effect. The dynamic stability was investigated by using the margin of stability (MoS), which was calculated as the difference between the anterior boundary of the base of support and the extrapolated position of the center of mass in the anterior-posterior direction.

MoS at recovery leg touchdown was significantly lower in the unexpected soft trials compared to the baseline, indicating a less stable posture. However, MoS increased ($p<0.05$) in both groups within the disturbed trials, indicating feedback adaptive improvements. Young and old participants showed differences in the handling of the perturbation in the course of several trials. The magnitude of the reactive adaptation after the fifth unexpected perturbation was significantly different to the first unexpected perturbation (o: 49±30%; y: 77±40%), showing a tendency ($p=0.065$) for higher values in the young participants.

Old individuals maintain the ability to adapt to feedback controlled perturbations. However, the locomotor behavior is more conservative compared to the young ones, leading to disadvantages in the reactive adaptation while disturbed walking.

5.2 Introduction

The incidence of falls for the elderly population increased and according with that the consequential fall-related injuries *(Blake et al. 1988; Tinetti et al. 1988; King & Tinetti 1995)*. Especially after unexpected perturbations like tripping, old subjects show a decreased recovery performance compared to young ones, leading to a higher occurrence of falls *(Thelen et al. 1997, 2000; Grabiner et al. 2005; Pijnappels et al. 2005; Karamanidis & Arampatzis, 2007)*. The reduced ability of the elderly population to regain balance can be attributed to an age-related decrease in muscle strength, tendon stiffness (*Schultz 1995; Grabiner et al. 2005; Karamanidis et al. 2008)* and a reduced rate of force development *(Vandervoort & McComas 1986)*. Furthermore, impairments in motor performance are not only dependent on changes in peripheral structures, but as well on central nervous system changes as age-related degeneration of motor cortical regions or neurotransmitter systems *(Seidler et al. 2010)*.

However, recovery performance can be modified by adaptations in a predictive and/or feedback based manner. Possible adjustments affect for example the magnitude of the base of support (prior as well as after the perturbation), the horizontal velocity of the center of mass and the position of the center of mass *(Bhatt et al. 2005, 2006 MacLellan & Patla 2006; Arampatzis et al. 2011)*. It is believed that supraspinal structures account for predictive locomotor adjustments *(Earhart et al. 2002a; Morton & Bastian 2006)* and determine the required feedforward motor control which is based on available knowledge about the movement. Humans are able to improve their dynamic stability in a predictive manner, counteracting an expected perturbation by ongoing adaptive adjustments based on prior experience *(Marigold & Patla 2002; Pai et al. 2003; van der Linden et al. 2007)*. Predictive adjustments may help to avoid or to reduce the consequence of perturbations. Reactive adjustments, on the other hand, depend on knowledge received during the movement and it is believed that spinal centers contribute to the reactive motor control *(Morton & Bastian 2006)*. With age, however, the quality of afferent sensory information as well as the capacity of the proprioceptive system decline (*Lord et al. 1996; Hurley et al. 1998; Patel et al. 2009)*. The reduced capacities of the proprioceptive and afferent system lead to higher sensory thresholds and less generated information about the disturbance. The altered feedback in turn may influence the response to unexpected perturbations.

Even though the feedback based corrections of the old adults after an unexpected perturbation during walking are less effective compared to the young ones *(Pijnappels et al. 2005; Bierbaum et al. 2010)*, the predictive adaptive improvements regarding the dynamic stability control are similar in both old and young adults *(Pavol et al. 2004; Bierbaum et al. 2010)*. Feedback based corrections to modified conditions rely to a great extent on the accuracy and correctness of the sensory information. Prior experience may accelerate the processing of given afferent information in the central nervous system, improving the feedback responses. In young participants, *van der Linden et al. (2007)* found a reduction in the EMG amplitudes of M. gastrocnemius medialis and M. tibialis anterior over several unexpected step down perturbations as compared to the first unexpected step down trial (i.e. perturbation without experience). This finding suggests a possible effect of prior experience of specific perturbations on reactive adjustments, however, age-related reactive adaptive improvements in dynamic stability control after perturbations during walking remains uninvestigated. Therefore the aim of this study was to investigate the adaptation of young and old persons with repeated unexpected perturbations during gait. We hypothesized that young as well as old adults will show adaptations in their reactive behavior to unexpected perturbations during locomotion, however, young adults would adapt faster and to a greater degree compared to the old ones.

5.3 Methods

5.3.1 Experimental design

Fourteen elderly (O: 67.3 ± 4.2 yrs.) and fourteen young (Y: 24.9 ± 2.4 yrs.) male subjects with similar body mass (O: 77.6 ± 7.5 kg; Y: 77.3 ± 7.5 kg) and body height (O: 173.4 ± 7.7 cm; Y: 178.5 ± 8.1 cm), all physically active and without neurological or musculoskeletal impairments, participated in the study after giving informed consent to the experimental procedure according to the rules of the local scientific board.

The subjects had to perform walking trials on a walkway (12 x 0.6 x 0.2 m^3) which included an exchangeable element (90 x 60 x 20 cm^3). This element was hidden with a cover sheet to be able to change the surface from hard to soft and vice versa without the knowledge of the subjects *(fig. 5-1)*.

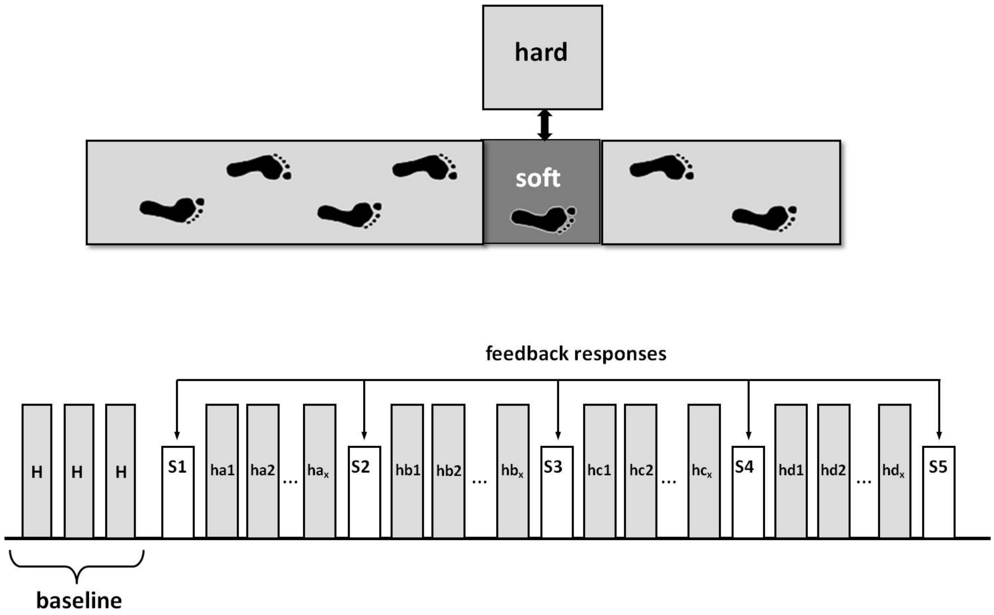

Figure 5-1: *Experimental protocol of hard and soft surface gait trials. The walkway included one covered exchangeable element which allowed changing the surface condition from hard to soft and vice versa without the knowledge of the participants. Three trials on hard surface presented the baseline and were followed by one unexpected soft surface trial (S1). Four to eight trials on hard surface followed S1 and all of the later soft trials to generate a "wash-out" of the experience of the soft surface.*

The soft element was made of foam with an upper surface consisting of relatively hard rubber material (depth = 0.8 cm). The deformation of the soft element during the walking trials was about 10 cm in depth, for both age groups, whereas the force deformation characteristic was non-linear. The hard element matched the walkways hardness. Although the subjects were informed that something in the walkway might change, they were unaware of the exact details. Thus they were unaware not only of the type of perturbation but also of the point in time when it might occur. The participants were told to not speak about the exact details of the protocol among each other. The protocol consisted of a total of about 28 gait trials, starting with three trials on the hard surface (baseline) and aimed at generating 5 unexpected, but similar perturbations of the gait. The number of trials all in all depended on the number of trials in-between the soft surface trials which were required to achieve the wash-out effect. Participants walked at 60% of their walk-to-run-velocity. Walk-to-run transition velocity (WRV) was determined for each participant on the basis of four to six walking trials before the main experiments. Both age groups achieved different maximal gait velocities ($young_{mean}$: 3.5 ± 0.33 m/s; old_{mean}: 3.19 ± 0.23 m/s) and therefore different target speed for the experimental trials. The speed of 60% of the WRV was chosen to obtain similar demands for the neuromuscular system of the two age-groups during the experiments *(Neptune & Sasaki 2004)*. The diverse walking speed and similar leg length led to a smaller Froude number for the elderly, indicating not only an absolute smaller walking speed but also a smaller walking speed relative to the leg length for the old compared to the young participants. The target velocity (60% of WRV) was controlled by light barriers and was predefined by a stick, which moved in the required speed in front of the subjects along the walkway. The speed and subsequently step length were maintained constant, in this manner, throughout the experimental trials so that the participants would always step with their right leg on the exchangeable element.

The experiment was designed to examine adaptive improvements of the participants after unexpected perturbations (i.e. reactive adaptive improvements) on the soft surface. We considered five soft trials in the entire experiment. The assessment of reactive adaptive improvements during the soft trials required unexpected perturbations. Therefore, the first soft trial was performed after three baseline trials on the hard surface without any information for the participants. Thus, it was an unexpected trial, executed without prior experience. In previous work we found predictive adjustments in expectance of the perturbation after the experience of the soft surface *(Bierbaum et al. 2010)*. In order to avoid this feedforward behavior, this first soft trial and the other four soft trials were followed by 4-10 trials on the hard surface – until no more predictive adjustments could be seen. We controlled this by the state of the dynamic stability (margin of stability, base of support etc.) prior to the perturbation. With this experimental design we were able to detect only unexpected perturbations on the soft trials and thus to assess the feedback responses of the examined participants.

5.3.2 Quantification of dynamic stability control

Ground reaction forces of the last step before the exchangeable element and the step on the exchangeable walkway element were collected at a sampling rate of 1080 Hz with two Kistler force plates (60 x 90 cm, Kistler, Winterthur, Switzerland) and were used to detect the touchdown. Kinematic data were recorded with the Vicon motion capture system (Model 624, Vicon, Oxford, UK) using 12 cameras operating at 120 Hz. Reflective markers were positioned at C7 and with four markers on a headband. For both body sides markers were fixed at the acromion, tuberositas radii, processus styloideus ulna, trochanter major femoris, knee joint space lateral, malleolis lateralis, tuber calcanei and at the second caput ossis metatarsis. Whole body kinematics were observed through smoothed marker trajectories *(Woltring filter routine, Woltring 1986)* and the center of mass was estimated on the basis of data from *Zatsiorsky & Seluyanov (1983).*

To quantify the dynamic stability we used the “extrapolated center of mass” concept formulated by *Hof et al. (Hof et al. 2005)*. The inverted pendulum model was used for the determination of dynamic stability. Margin of stability as a criterion for the state of stability of the human body was calculated as follows:

$$b_{x} = u_{max} - CM_{exp}$$

where b_x indicates the margin of stability in the anterior-posterior direction, u_{max} is the anterior boundary of the base of support and CM_{exp} is the position of the extrapolated center of mass in the anterior-posterior direction. CM_{exp} is calculated from the horizontal (anteroposterior) component of the projection of the center of mass (CM) to the ground (P_{XCM}), the horizontal CM velocity (V_{XCM}), the acceleration of gravity (g) and the distance between the CM and the center of the ankle joint in the sagittal plane (l) in the following way:

$$CM_{\exp} = P_{XCM} + \frac{V_{XCM}}{\sqrt{g/l}}.$$

The positions of the foot markers at the calcanei and the 2nd metatarsal bone were registered relative to individual sketches of the subjects´ feet so that the boundaries of the base of support could be calculated out of the kinematic data. Base of support refers to the distance from the anterior boundary of the leading foot to the posterior boundary of the rear foot. The human body shows a stable position if the location of the CM_{exp} is within the boundaries of the base of support. Negative values of margin of stability indicate an unstable position of the body and therefore additional actions need to be performed to maintain the balance. Margin of stability *(fig. 5-2)* was determined in the antero-posterior direction since the extrapolated center of mass mainly shifts in the anterior direction after the perturbation due to the increased velocity of the center of mass. These parameters were used to quantify the dynamic stability control while walking and were analyzed at two distinct time points: a) touchdown of the disturbed leg (before the perturbation, TDdist), b) touchdown of the recovery leg (TDrec). Disturbed leg refers to the leg, which steps on the exchangeable element and the recovery leg is defined as the leg, which helps to regain balance after the perturbation.

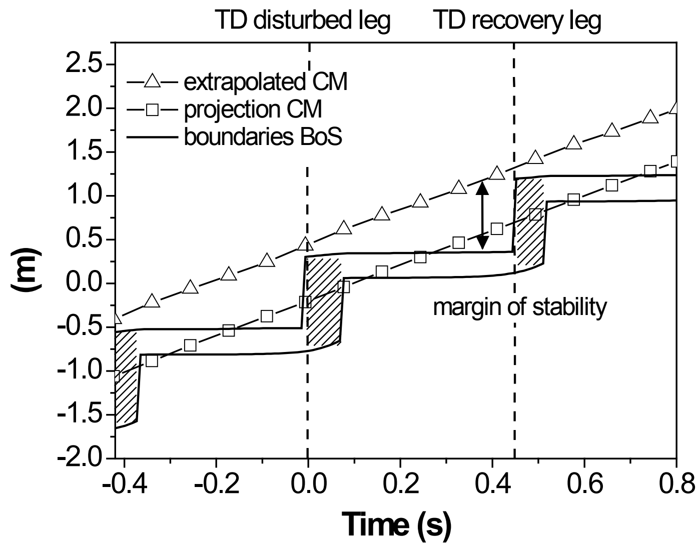

Figure 5-2: *Parameters of dynamic stability control according to Hof et al. (2005), using the example of one baseline trial: extrapolated center of mass (CM), projection of CM to the ground and boundaries of the base of support (BoS) in the anterior-posterior direction. The top solid line shows the anterior boundary and the bottom line the posterior boundary of the BoS. The shaded area indicates the double stance phase. Note that in this specific walking velocity (~1.8 m/s) the extrapolated CM is located permanently outside of the base of support (i.e. negative values of the margin of stability).*

The magnitude of the reactive adaptation in dynamic stability during the experimental protocol has been calculated as follows:

$$Adaptation = (1 - (〖MoS〗\_S5 - 〖MoS〗\_B)/(〖MoS〗\_S1 - 〖MoS〗\_B))) \text{ x } 100$$

where S1 and S5 indicate first and fifth unexpected soft surface trial respectively. The mean values of the three baseline trials on hard surface were used to establish the base level which is represented by B. Margin of stability at TDrec of these trials was used for the calculation of the adaptation.

5.3.3 Statistics

A repeated measurement ANOVA with the different experimental trials as intersubject variables and the age-group as the between-subject variable was performed to examine the trial related differences in the analyzed dynamic stability parameters (margin of stability, base of support, position of the extrapolated CM, horizontal component of the projection of the CM to the ground, horizontal CM-velocity, the term $\sqrt{g/l}$, swing time of the recovery leg and first local maximum of the horizontal CM-velocity). The difference in the magnitude of the adaptation of the margin of stability was analyzed by a two-way ANOVA with the fixed variables age-group (two levels: old and young) and repetition (two levels: trial 1 vs. trial 5). Anthropometric data and the magnitude of the reactive adaptation were compared with a t-test for non-dependent samples. The level of significance for all statistical comparisons was set to α= 0.05.

5.4 Results

To assess the “wash-out” effect in our experiment, we compared the horizontal velocity and the margin of stability at touchdown of the disturbed leg between baseline trials and the soft surface (disturbed) trials. Comparing baseline and soft trials, both age groups showed no differences in the horizontal velocity of center of mass (CM) as well as in the margin of stability at touchdown of the disturbed leg (TDdist), indicating similar stability at the time of the perturbation *(table 5-1)*.

Table 5-1: *Means ± SD of the margin of stability and the horizontal velocity of the CM for the young and elderly people at touchdown of the disturbed leg at baseline and the following experimental soft trials.*

	Margin of stability [cm]		Horizontal CM-velocity [m/s]	
	Young	**Old**	**Young**	**Old**
Baseline	-7.59 ± *6.70*	-4.90 ± *4.58*	2.18 ± *0.22*	2.01 ± *0.14*
S1-trial	-7.81 ± *5.72*	-6.87 ± *5.26*	2.19 ± *0.20*	2.05 ± *0.15*
S2-trial	-8.03 ± *7.22*	-3.92 ± *6.38*	2.20 ± *0.24*	2.00 ± *0.17*
S3-trial	-6.26 ± *6.00*	-3.11 ± *5.70*	2.17 ± *0.20*	2.00 ± *0.15*
S4-trial	-6.62 ± *6.42*	-3.95 ± *5.55*	2.16 ± *0.21*	2.04 ± *0.15*
S5-trial	-6.48 ± *3.16*	-2.81 ± *3.16*	2.15 ± *0.20*	2.01 ± *0.16*

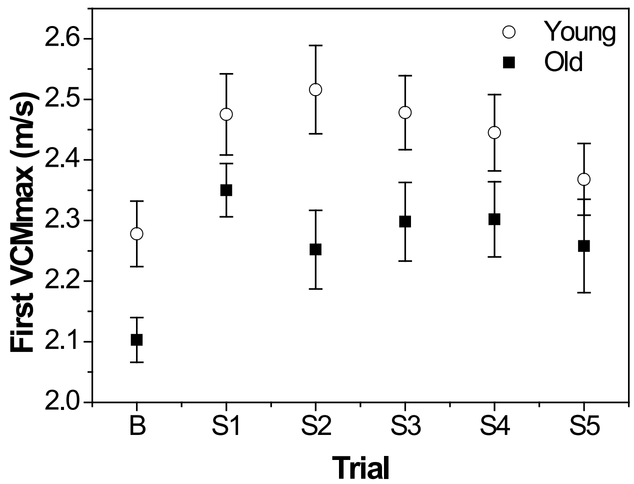

Figure 5-3: *Mean values and standard error of mean of the first local maximum velocity of center of mass in the horizontal direction for young and old participants after the touchdown of the disturbed leg. Note the statistically significant differences to baseline for all soft surface trials (p<0.05), which indicates the effect of the soft surface on the center of mass behavior. Differences between the age groups originate in the different walking velocities.*

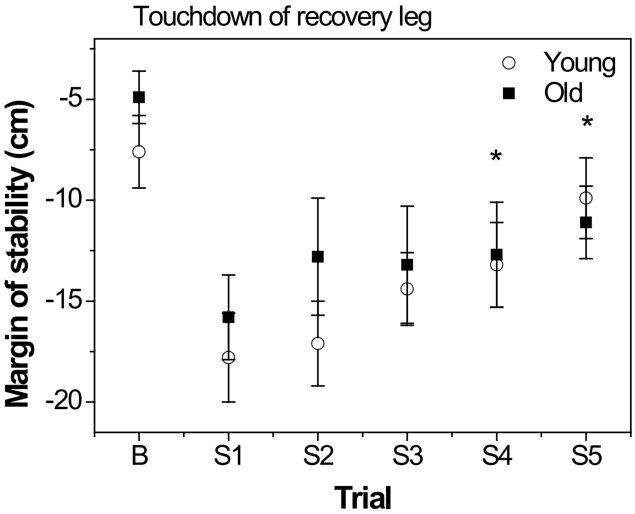

Figure 5-4: *Mean values and standard error of mean for margin of stability at touchdown of the recovery leg in the baseline (B) and the unexpected soft trials (S1-S5). Note that all soft trials showed lower values in the margin of stability compared to the baseline (p<0.05). *: statistically significant difference as compared to the first unexpected soft surface (S1) (p<0.05).*

Furthermore, although the first local maximum of the horizontal CM-velocity is higher in all soft trials compared to the baseline walking trials *(fig. 5-3)*, this maximum did not show any significant ($p>0.05$) differences in-between the soft trials, indicating similar acute consequences of the perturbation within the disturbed trials. Differences between the age groups originated from the differing walking velocities.

Compared to the baseline, margin of stability decreased for both age groups in the first unexpected soft surface trial at touchdown of the recovery leg (TDrec), reflecting a more unstable position at touchdown as a consequence of the experienced perturbation *(fig. 5-4)*. In the following four soft surface trials the margin of stability increased continuously for both age groups compared to the first soft trial, resulting in a statistically significant ($p<0.05$) increase in the trials S4 and S5. The base of support at touchdown of the recovery leg showed in both age groups higher values ($p<0.05$) for the soft trials compared to the baseline but not ($p<0.05$) in-between the soft trials *(fig. 5-5a)*. The extrapolated CM at TDrec showed a more anterior position ($p<0.05$) in both groups during the soft trials compared to the baseline. Within the soft trials, young participants reduced the anterior-posterior position of the extrapolated center of mass continuously and attained a significantly ($p<0.05$) lesser anterior position of the extrapolated CM at 5th as compared to the 1st trial *(fig. 5-5b)*. The horizontal component of the projected CM to the ground showed the same development as the position of the extrapolated CM. Both groups moved the CM ($p<0.05$) during the disturbed trials to a more anterior position compared to the baseline *(fig. 5-6)*. Within the soft trials only the young participants demonstrated a significantly decreased ($p<0.05$) projected CM at trial 5 compared to trial 1. Values for the term $\sqrt{g/l}$ and the horizontal velocity of CM at touchdown of the recovery leg showed no differences for the later soft trials in comparison to the first unexpected soft surface for both age groups *(table 5-2)*. The duration of the swing phase of the recovery leg within the soft trials increased significantly ($p<0.05$) for the young participants compared to the baseline *(fig. 5-7)*. However, within the soft (disturbed) trials, young participants reduced the duration of the swing phase ($p<0.05$).

Both groups showed a significant ($p<0.05$) adaptive improvement in their stability, visible in the magnitude of the adaptation in the margin of stability at touchdown of the recovery leg at the last soft trial (trial 5) *(fig. 5-8)*. Although not significant, there were trends indicating that the young participants showed a tendency ($p=0.065$) for higher reactive adaptive improvements as compared to the old.

Table 5-2: *Parameters of dynamic stability control at touchdown of the recovery leg. Means ± SD of the term (* $\sqrt{g/l}$ *: eigenfrequency of a pendulum with length l and the acceleration of gravity g) and the horizontal velocity of the CM for the young and elderly people at touchdown of the recovery leg at baseline and the following experimental soft trials.*

	Term [$^{-1}$]		**Horizontal CM-velocity [m/s]**	
	Young	**Old**	**Young**	**Old**
Baseline	3.23 ± *0.16*	3.23 ± *0.06*	2.18 ± *0.22*	2.01 ± *0.14*
S1-trial	3.14 ± *0.07*	3.17 ± *0.07*	2.45 ± *0.24*	2.34 ± *0.16*
S2-trial	3.16 ± *0.08*	3.18 ± *0.08*	2.51 ± *0.28*	2.23 ± *0.24*
S3-trial	3.15 ± *0.08*	3.16 ± *0.07*	2.47 ± *0.24*	2.30 ± *0.21*
S4-trial	3.16 ± *0.09*	3.18 ± *0.07*	2.44 ± *0.23*	2.30 ± *0.20*
S5-trial	3.17 ± *0.07*	3.16 ± *0.07*	2.35 ± *0.21*	2.26 ± *0.21*

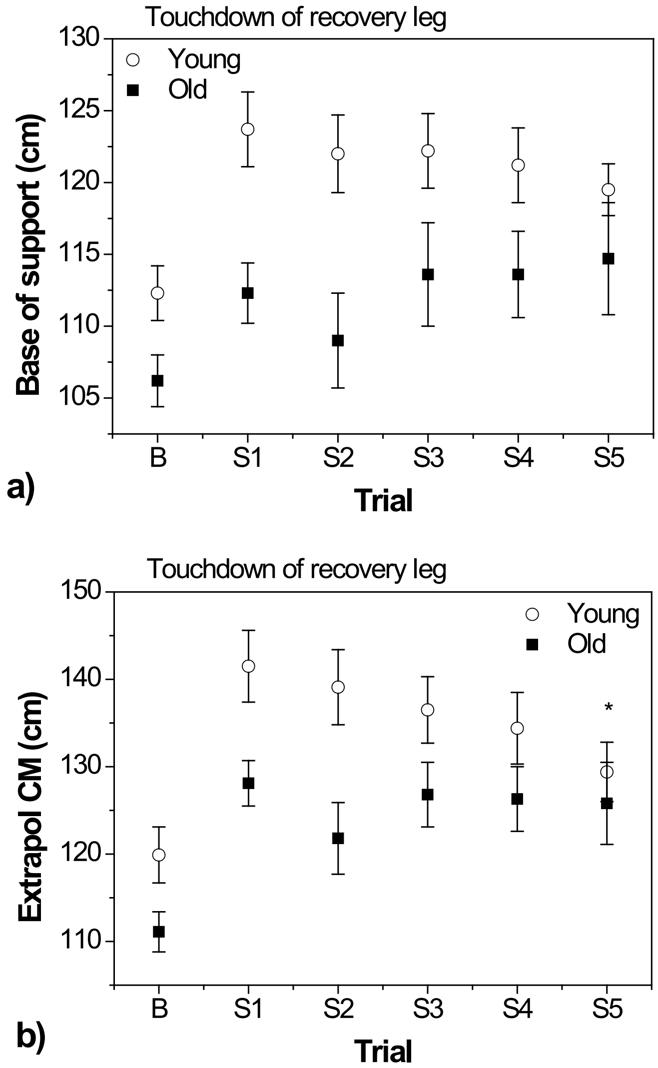

Figure 5-5: *Mean and standard error of mean of (a) base of support and (b) extrapolated center of mass at touchdown of the recovery leg. Note that the base of support as well as the extrapolated center of mass demonstrated higher values for the soft trials compared to the baseline (p<0.05).*
**: statistically significant difference to the first unexpected soft surface (S1) (p<0.05).*

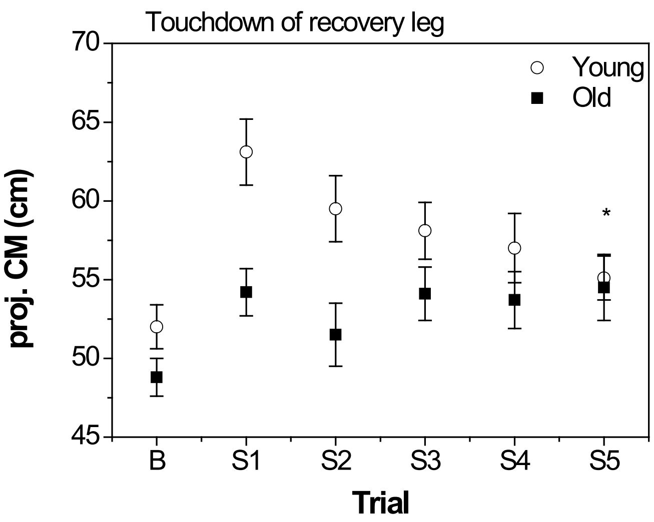

Figure 5-6: *Mean and standard error of mean of the horizontal component of the projected center of mass to the ground at touchdown of the recovery leg. Note that all soft trials showed higher values in the horizontal component of the projected CM to the ground compared to the baseline (p<0.05). *: statistically significant difference to the first unexpected soft surface (S1) (p<0.05).*

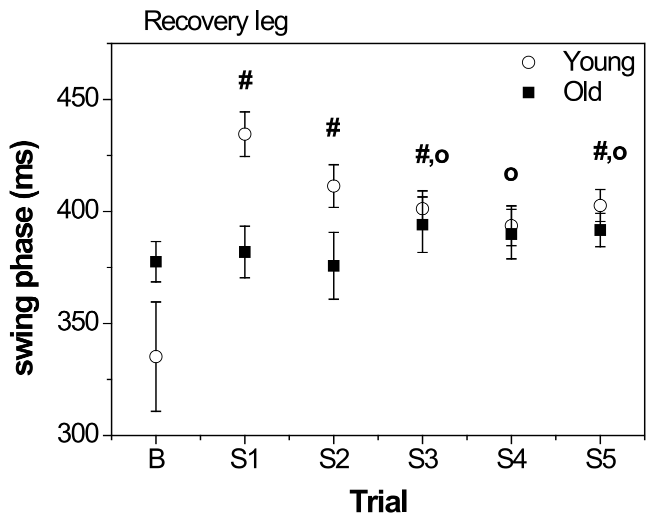

Figure 5-7: *Mean and standard error of mean of the duration of the swing phase of the recovery leg. #: statistically significant difference to baseline for the young participants (p<0.05). o: statistically significant difference to the first unexpected soft surface (S1) for the young participants (p<0.05).*

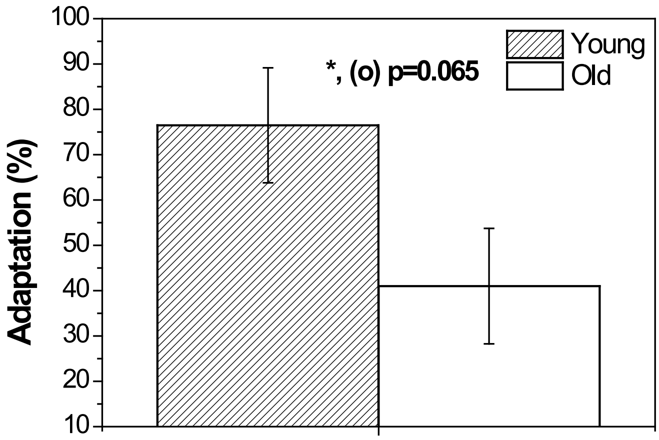

Figure 5-8: *Mean and standard error of mean of the adaptation in the margin of stability at touchdown of the recovery leg. * stat. sign. diff. to zero values (p < 0.05); (°): tendency to statistically sign. differences between age groups (p=0.065).*

5.5 Discussion

This study investigated possible effects of prior experience on reactive behavior with respect to age. We examined in what way young and old male individuals adapt their reactive behavior to repetitive unexpected gait perturbations with respect to dynamic stability. We hypothesized that young as well as old adults will show reactive adaptations to unexpected perturbations during locomotion and that young adults will adapt to a greater degree compared to the old ones.

In the course of several unexpected perturbations of the gait, both age groups showed significant adaptive improvements in the dynamic stability after the perturbation, confirming the first hypothesis. This result may be a key towards designing new fall prevention strategies for the elderly population as it shows that the experience of simulated perturbations during walking improves the reactive behavior regarding stability control. However, although the second hypothesis has not been clearly confirmed (young participants showed only a tendency for a greater reactive adaptation after the fifth perturbation compared to the old ones), it does indicate a trend suggesting that the locomotor behavior among the old participants with respect to dynamic stability control after unexpected perturbations is less variable than that of the young ones, leading to disadvantages for the elderly in the reactive adaptation while disturbed walking. The reason for the reactive adaptational improvements of the young adults in-between the unexpected soft trials was not an increase in base of support (base of support remained indifferent within the unexpected trials) but rather a faster recovery step. The duration of the swing phase of the recovery leg decreased significantly for the young subjects from the first to the fifth unexpected trial leading to a more posterior position of the center of mass at touchdown of the recovery leg compared to the first unexpected perturbation. By keeping the base of support similar and decreasing the duration of the swing phase during the perturbed trials, young participants performed a faster step to the front after experiencing the perturbation several times. The positioning of the CM more posteriorly implied that the young participants placed their extrapolated center of mass posteriorly - leading to a higher margin of stability after the perturbation (i.e. higher stability) in the last two unexpected trials.

The old adults did not change significantly any of the components of dynamic stability within the five unexpected trials. The significant increase in margin of stability (i.e. improved stability) at touchdown of the recovery leg in the last two trials compared to the baseline happened by a combination of a non-significantly modified base of support (i.e. increase) and horizontal CM velocity (i.e. decrease). Old participants remained in their early developed pattern until the last unexpected perturbation. No large modifications of the first applied behavior during the course of several perturbations could be seen. There was a trend ($p=0.065$) suggesting greater reactive locomotor adaptability among the young participants, which is due to their increased readiness to adopt a new locomotor behavior. However, despite the less variable locomotor characteristic of the old compared to the young participants, the age-related biological impairments do not inhibit the formation of reactive adaptive improvements in the dynamic stability. These findings may have important implications for the quality of life in the elderly population. The reduced ability of old participants to modify drastically their locomotor behavior after repeated unexpected perturbations may contribute to their increased risk of falling during daily activities. On the other hand, the retained ability to generate reactive adaptive improvements based on prior experience after unexpected perturbations gives evidence, that old adults may increase their ability to regain balance after disturbances.

The improved handling of perturbations can be attained by predictive adjustments prior to the perturbation. Modifications of the gait and postural behavior in anticipation of possible perturbations could affect the effectiveness of the reactive responses positively. It is well known that predictive adjustments to repeated perturbations reduce significantly the consequences of the expected perturbation *(Marigold & Patla 2002; Bhatt et al. 2006; Bierbaum et al. 2010)*. Therefore, participants performed several trials on hard surface after each unexpected soft trial to achieve a normal gait pattern prior to the next perturbation. This was done to attain a wash-out effect of the previous experienced perturbation. The effectiveness of the wash-out phase was investigated by the determination of the components of dynamic stability prior to the perturbation. Neither young nor old participants showed modified components of dynamic stability immediately before the perturbation compared to baseline values. Furthermore, the acute consequences of the perturbation (measured by the first local maximum of the horizontal CM-velocity) remained similar for all soft trials, indicating no influence of possible predictive adjustments. Therefore, the found locomotor improvements in dynamic stability after the disturbance are associated with an enhancement of the reactive responses.

In conclusion, young individuals tended to show higher reactive adaptational improvements regarding the dynamic stability control compared to the old ones. The reason was a modification in the increase of base of support (i.e. faster increase of base of support) after the repeated unexpected perturbations. Elderly participants also demonstrated an improvement in dynamic stability after unexpected perturbations while walking, however by less modifications of their first applied behavior. These results could help design future fall prevention strategies which should consider the parameters of dynamic stability and especially the mechanism of increasing the base of support.

Conflict of interest statement

The authors have no conflicts of interest to disclose.

5.6 References

Arampatzis, A., Peper, A., and Bierbaum, S. (2011). Exercise of mechanisms for dynamic stability control increases stability performance in the elderly. J Biomech *44*, 52–58.

Bhatt, T., Wening, J.D., and Pai, Y.-C. (2005). Influence of gait speed on stability: recovery from anterior slips and compensatory stepping. Gait & Posture *21*, 146–156.

Bhatt, T., Wening, J.D., and Pai, Y.-C. (2006). Adaptive control of gait stability in reducing slip-related backward loss of balance. Exp Brain Res *170*, 61–73.

Bierbaum, S., Peper, A., Karamanidis, K., and Arampatzis, A. (2010). Adaptational responses in dynamic stability during disturbed walking in the elderly. J Biomech *43*, 2362–2368.

Blake, A.J., Morgan, K., Bendall, M.J., Dallosso, H., Ebrahim, S.B., Arie, T.H., Fentem, P.H., and Bassey, E.J. (1988). Falls by elderly people at home: prevalence and associated factors. Age Ageing *17*, 365–372.

Earhart, G.M., Fletcher, W.A., Horak, F.B., Block, E.W., Weber, K.D., Suchowersky, O., and Melvill Jones, G. (2002a). Does the cerebellum play a role in podokinetic adaptation? Exp Brain Res *146*, 538–542.

Grabiner, M.D., Owings, T.M., and Pavol, M.J. (2005). Lower extremity strength plays only a small role in determining the maximum recoverable lean angle in older adults. J Gerontol A Biol Sci Med Sci *60*, 1447–1450.

Hof, A.L., Gazendam, M.G.J., and Sinke, W.E. (2005). The condition for dynamic stability. J Biomech *38*, 1–8.

Hurley, M.V., Rees, J., and Newham, D.J. (1998). Quadriceps function, proprioceptive acuity and functional performance in healthy young, middle-aged and elderly subjects. Age Ageing *27*, 55–62.

Karamanidis, K., and Arampatzis, A. (2007). Age-related degeneration in leg-extensor muscle-tendon units decreases recovery performance after a forward fall: compensation with running experience. Eur J Appl Physiol *99*, 73–85.

Karamanidis, K., Arampatzis, A., and Mademli, L. (2008). Age-related deficit in dynamic stability control after forward falls is affected by muscle strength and tendon stiffness. J Electromyogr Kinesiol *18*, 980–989.

King, M.B., and Tinetti, M.E. (1995). Falls in community-dwelling older persons. J Am Geriatr Soc *43*, 1146–1154.

van der Linden, M.H., Hendricks, H.T., Bloem, B.R., and Duysens, J. (2009). Hitting a support surface at unexpected height during walking induces loading transients. Gait & Posture *29*, 255–260.

Lord, S.R., Lloyd, D.G., and Li, S.K. (1996). Sensori-motor function, gait patterns and falls in community-dwelling women. Age Ageing *25*, 292–299.

MacLellan, M.J., and Patla, A.E. (2006). Adaptations of walking pattern on a compliant surface to regulate dynamic stability. Exp Brain Res *173*, 521–530.

Marigold, D.S., and Patla, A.E. (2002). Strategies for dynamic stability during locomotion on a slippery surface: effects of prior experience and knowledge. J Neurophysiol *88*, 339–353.

Morton, S.M., and Bastian, A.J. (2006). Cerebellar contributions to locomotor adaptations during splitbelt treadmill walking. J Neurosci *26*, 9107–9116.

Neptune, R.R., and Sasaki, K. (2005). Ankle plantar flexor force production is an important determinant of the preferred walk-to-run transition speed. J Exp Biol *208*, 799–808.

Pai, Y.-C., Wening, J.D., Runtz, E.F., Iqbal, K., and Pavol, M.J. (2003). Role of feedforward control of movement stability in reducing slip-related balance loss and falls among older adults. J Neurophysiol *90*, 755–762.

Patel, M., Magnusson, M., Kristinsdottir, E., and Fransson, P.-A. (2009). The contribution of mechanoreceptive sensation on stability and adaptation in the young and elderly. Eur J Appl Physiol *105*, 167–173.

Pavol, M.J., Runtz, E.F., and Pai, Y.-C. (2004). Young and older adults exhibit proactive and reactive adaptations to repeated slip exposure. J Gerontol A Biol Sci Med Sci *59*, 494–502.

Pijnappels, M., Bobbert, M.F., and Dieën, J.H. van (2005). Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. Gait & Posture *21*, 388–394.

Schultz, A.B. (1995). Muscle function and mobility biomechanics in the elderly: an overview of some recent research. J Gerontol A Biol Sci Med Sci *50 Spec No*, 60–63.

Seidler, R.D., Bernard, J.A., Burutolu, T.B., Fling, B.W., Gordon, M.T., Gwin, J.T., Kwak, Y., and Lipps, D.B. (2010). Motor control and aging: links to age-related brain structural, functional, and biochemical effects. Neurosci Biobehav Rev *34*, 721–733.

Thelen, D.G., Muriuki, M., James, J., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (2000). Muscle activities used by young and old adults when stepping to regain balance during a forward fall. J Electromyogr Kinesiol *10*, 93–101.

Thelen, D.G., Wojcik, L.A., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (1997). Age differences in using a rapid step to regain balance during a forward fall. J Gerontol A Biol Sci Med Sci *52*, M8–13.

Tinetti, M.E., Speechley, M., and Ginter, S.F. (1988). Risk factors for falls among elderly persons living in the community. N Engl J Med *319*, 1701–1707.

Vandervoort, A.A., and McComas, A.J. (1986). Contractile changes in opposing muscles of the human ankle joint with aging. J Appl Physiol *61*, 361–367.

Woltring, H.J. (1986). A Fortran package for generalized, cross-validatory spline smoothing and differentiation. Adv Eng Software (1978) *8*, 104–113.

Zatsiorsky, V.M., and Seluyanov, V. (1983). The mass and inertial characteristics of main segments of the human body. In Biomechanics VIII-B, (Champaign, IL: Human Kinetics Publishers), pp. 1152–1159.

6. Third study: Exercise of mechanisms of dynamic stability improves the stability state after an unexpected gait perturbation in elderly

Stefanie Bierbaum[1,2]
Andreas Peper[1]
Adamantios Arampatzis[1,3]

[1] Department of Training and Movement Sciences, Humboldt-University Berlin

[2] Motor Performance and Cognition Laboratory, University of Stuttgart

[3] Center for Sports Science and Sports Medicine Berlin

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6.1 Abstract

Unexpected changes during gait challenge elderly individuals to a greater degree than young adults. However, the adaptive potential of elderly seems to be retained and therefore the training of the mechanisms of dynamic stability as well as muscle strength training may improve the dynamic stability after unexpected perturbations. 38 subjects (65-75 years) participated in the study, divided into 2 experimental groups (stability training group, ST: n=14; mixed training group, MT: n=14) and a control group (CG: n=10). Both experimental groups performed exercises which focused on the mechanisms of dynamic stability. Additionally, the MT-group executed a training to improve muscle strength. Session volume and duration was equal for both groups (14 weeks, twice a week, ~1.5h per session). Pre- and post-intervention, subjects performed a gait protocol with an induced unexpected perturbation. Post-intervention, the margin of stability was significantly increased after the unexpected perturbation in the ST-group, indicating an improvement in stability state (pre: -30.3±5.9cm; post: -24.1±5.2cm). Further, both intervention groups increased their base of support after the intervention to regain balance after gait perturbation, whereas only the ST-group showed a statistically significant improvement (ST_{pre}: 90.9±6.6cm, ST_{post}: 98.2±8.5cm; MT_{pre}: 91.4±6.2cm, MT_{post}: 97.9±12.7cm). The CG showed no differences between pre- and post-measurements. The exercise of the mechanisms of dynamic stability led to a better application of these mechanisms after an unexpected perturbation during gait. We suggest that the repeated exercise of the mechanisms of dynamic stability contributes to significant improvements in postural stability. Additional strength training for healthy elderly individuals, however, shows no further effect on the ability to recover balance after unexpected perturbations during gait.

6.2 Introduction

Human biped locomotion represents a major challenge to the postural system, especially in the elderly, whereas postural stability constitutes the required ability to maintain equilibrium under various static and dynamic conditions *(Horak 2006; Shumway-Cook & Woollacott 2006).* Individuals have increasing problems with balance and tend to fall more often as they grow older *(Lord et al. 1993)*. Therefore, fall-related injuries in the elderly population are an important economic as well as social problem *(Blake et al. 1988; Tinetti et al. 1988; Etman et al. 2012).* In Germany, for example, hip fractures of nursing home residents are estimated to cost about 8160 Euros each, and the overall costs of falls amongst the elderly account for about 2.1 – 3.4 billion Euros per year *(Heinrich et al. 2012).*

There are several factors which contribute to an increased fall risk in the elderly population; one of those factors is the experience of unexpected perturbations. Elderly individuals are less able to handle sudden, unexpected changes *(Pijnappels et al. 2005; Karamanidis & Arampatzis 2007; Bierbaum et al. 2010)*. Reasons for the decreased ability to recover balance are the age-related reduction in muscle-strength and tendon stiffness *(Grabiner et al. 2005; Karamanidis et al. 2008)*, the delayed generation of propulsive ground reaction forces and joint torques *(Robinovitch et al. 2002; Pijnappels et al. 2005; Tseng et al. 2009),* and the lower muscular-contraction velocities *(Hortobagyi et al. 1995; Thelen et al. 1997)*. If the postural system is unstable, stepping strategies seem to be less successful for the elderly due to reductions in step length and speed *(Wojcik et al. 2001; Karamanidis et al. 2008; Bierbaum et al. 2010)*. Furthermore, decreased recovery performance is attributed to deficits in the application of the mechanisms of dynamic stability *(Arampatzis et al. 2008).*

However, the control of dynamic stability during locomotion is not a fixed process, but it can adapt to different situations and be modified through exercise. Earlier studies have shown that elderly participants are able to adapt to repeated locomotor perturbations in a predictive as well as in a reactive manner *(Bhatt et al. 2006; Heiden et al. 2006; Bierbaum et al. 2010, 2011)*. These studies demonstrate that locomotor adaptability is still preserved in the elderly. However, although the predictive adjustments of the elderly participants during gait are similar to that of younger ones *(Bierbaum et al.*

2010), elderly participants showed a somewhat lower adaptive reactive potential after unexpected perturbations *(Bierbaum et al. 2011)*. Therefore, we conclude that the main deficits of stability performance in elderly are the reactive responses after unexpected perturbations during walking *(Pijnappels et al. 2005; Bierbaum et al. 2010, 2011)*. The adaptation potential of the old adults is very important in fall prevention, because adequate interventions, aiming to improve the stability performance in the elderly population, may reduce the risk of falls. From a practical point of view, it is impossible for individuals to identify and practice all possible specific perturbations which could lead to falls during daily life. Therefore, it is important to identify underlying mechanisms of dynamic stability which are relevant for the maintenance of the postural stability after gait perturbations. A more general exercise intervention including those mechanisms could allow elderly people to apply the mechanisms in different situations and, therefore, help them to regain balance.

From a biomechanical point of view, there are three mechanisms responsible for maintaining postural stability after perturbations: (a) increase in the base of support, (b) counter-rotating segments around the center of mass (CoM) and (c) the application of an external force (not the ground reaction force) *(Hof et al. 2005)*. A recent publication of our group *(Arampatzis et al. 2011)* showed that exercise of the mechanisms responsible for dynamic stability increased the stability performance of elderly adults (~30%) after a simulated forward fall. However, most falls in the elderly population happen during walking *(Rubenstein 2006).* To the best of our knowledge, no systematic study has been performed to examine the effect of training of the mechanisms of dynamic stability on the recovery performance after an unexpected perturbation during walking in elderly adults. Therefore, elderly participants were trained with a training protocol of various exercise tasks, which integrated the mechanisms responsible for dynamic stability *(see Arampatzis et al. 2011)*. We hypothesized that the exercise of the mechanisms of dynamic stability would contribute to an improvement of the stability performance after an unexpected gait perturbation in the elderly, indicating a transfer of the intervention into diverse situations. However, systematic reviews about the impact of intervention programs on fall risk reveal that multifactorial programs are more effective to reduce the number of falls *(Feder et al. 2000; Gillespie et al. 2010)*. Muscle strength training has shown not only to increase muscle strength but also muscle power and the ability to neurally activate motor units

(Hunter et al. 2004). Therefore, we hypothesized that a combined intervention program including exercising the mechanisms of dynamic stability and exercising muscle strength would show a higher improvement in the recovery performance of older adults. Accordingly, the purpose of this study was to examine the effects of specifically exercising the mechanisms of dynamic stability on the ability of elderly adults to regain balance after unexpected perturbations during walking. Furthermore, we wanted to examine the additional effect of increased muscle strength.

6.3 Methods

6.3.1 Participants

We examined elderly, non-sport active adults, aged 65 to 75 years. Seventy five healthy individuals gave their informed consent to the experimental procedure according to the rules of the Declaration of Helsinki. Exclusion criteria included serious neuromuscular or musculoskeletal impairments and any lower limb pain at the ankle, knee, or hip joint. Further exclusion criteria were: any medication during the study (e.g. against joint pain), any history of major trauma or major systemic diseases. Participants were randomly assigned to two experimental groups and one control group. Thirty-eight subjects finished the whole experimental design, whereas 14 participants were included in the stability training group (ST; 10 females and 4 males) and 14 in the mixed training group (MT; 10 females and 4 males). The control group consisted of ten persons (CG; five females and 5 males) and did not perform any exercise program *(fig. 6-1)*.

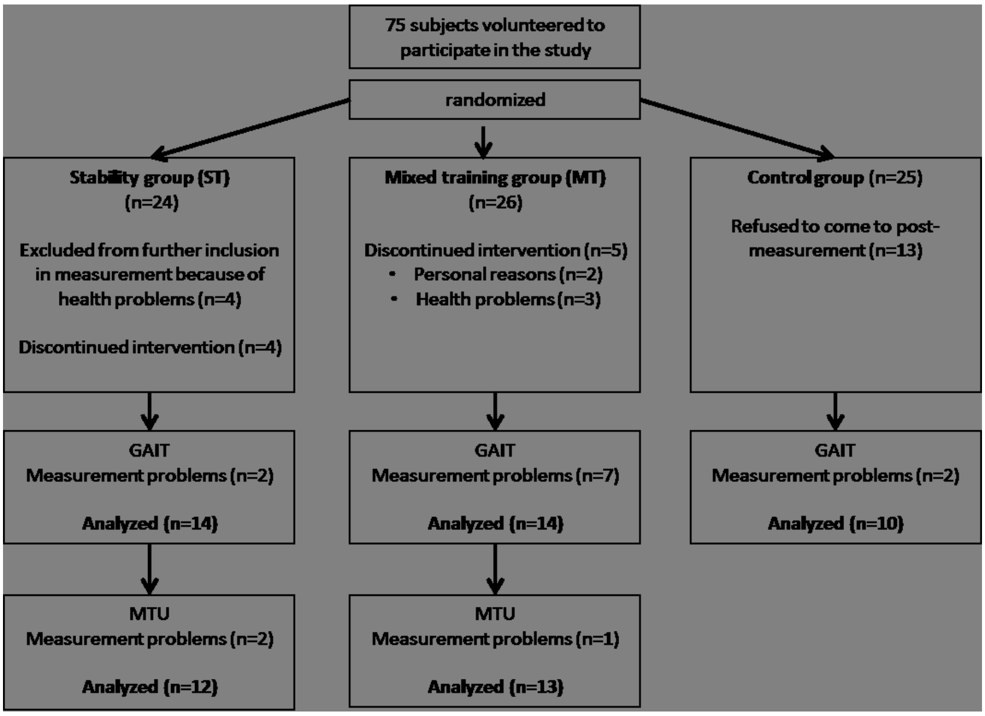

Figure 6-1: *Flow chart of recruitment and participation in the study. The performed gait measurement is named as "GAIT". "MTU" refers to the strength measurements of the muscle tendon unit (knee extensors and ankle plantarflexors).*

6.3.2 Exercise program

Both intervention groups exercised twice a week, 1.5h per session for 14 weeks. The ST-group performed exercise tasks including mechanisms of dynamic stability. The MT-group also performed such exercises, but additionally accomplished a strength-training program for the lower extremities.

Exercise tasks for the training of dynamic stability included two mechanisms of dynamic stability *(i.e. increase of the base of support & counter-rotating segments around the center of mass, see also Arampatzis et al. 2011).* The mechanism "increase of the base of support" was practiced with the following exercises: large and small, fast and slow, single and multiple steps in anterior-posterior and medio-lateral direction in order to stabilize the body in different positions. Furthermore, participants were supposed to use large, compensatory steps to regain postural stability after perturbations. According to *Maki and McIlroy (1997)*, compensatory steps are characterized by their reactive nature, an absence of functional anticipatory control, early initiation and rapid execution and possible adaptive changes in consequence of postural perturbations. The mechanism "counter-rotating segments around the center of mass" was included in the training of arm and leg movements (no compensatory steps) during standing and walking to maintain balance. Participants exercised on narrow support surfaces such as beams, thick ropes, or small bars, as well as on several compliant surfaces. Counter-rotations were initiated by walking with a small base of support, hopping, or landing. Perturbations for the training of both mechanisms were induced by catching or throwing balls during standing or walking on a normal or constricted walkway, by external forces, and by the use of an unstable oscillatory platform on which the participants had to maintain postural stability in a one- or two-legged stance. The difficulty of all exercises was adapted to each individual's ability *(see also Chodzko-Zajko et al. 2009)*. The ST-group performed those exercises including mechanisms of dynamic stability for 1.5 h and the MT-group for about 45 min per session. For the remaining 45 min the MT-group performed strength exercises using strength training machines: knee extension and flexion, hip flexion, and ankle extension. The strength training included three sets of 10 to 15 repetitions at 50-70% of their one-repetition maximum. The intervention program was always supervised by two experienced sport scientists.

6.3.3 Evaluation

A gait protocol on a gangway was performed before and after the 14 weeks of intervention. The gangway (12m x 0.6m x 0.2m) included one exchangeable element (0.9m x 0.6m x 0.2m), which was hidden by a cover sheet to be able to change the surface from hard to soft without the knowledge of the participants *(fig. 6-2)*. The soft element was made of foam with an upper surface consisting of relatively hard rubber material (depth=0.8cm). The deformation of the soft element during the walking trials was about 10cm in depth for all groups, whereas the force deformation characteristic was non-linear. The hard element consisted of the same material as the gangway. To attain the target position of the right leg (middle of the exchangeable element), the participants were advised to walk in a predefined velocity throughout the experimental trials whereas the practitioners adjusted the starting position if required. Trials with incorrect placement of the right foot were excluded from further analysis.

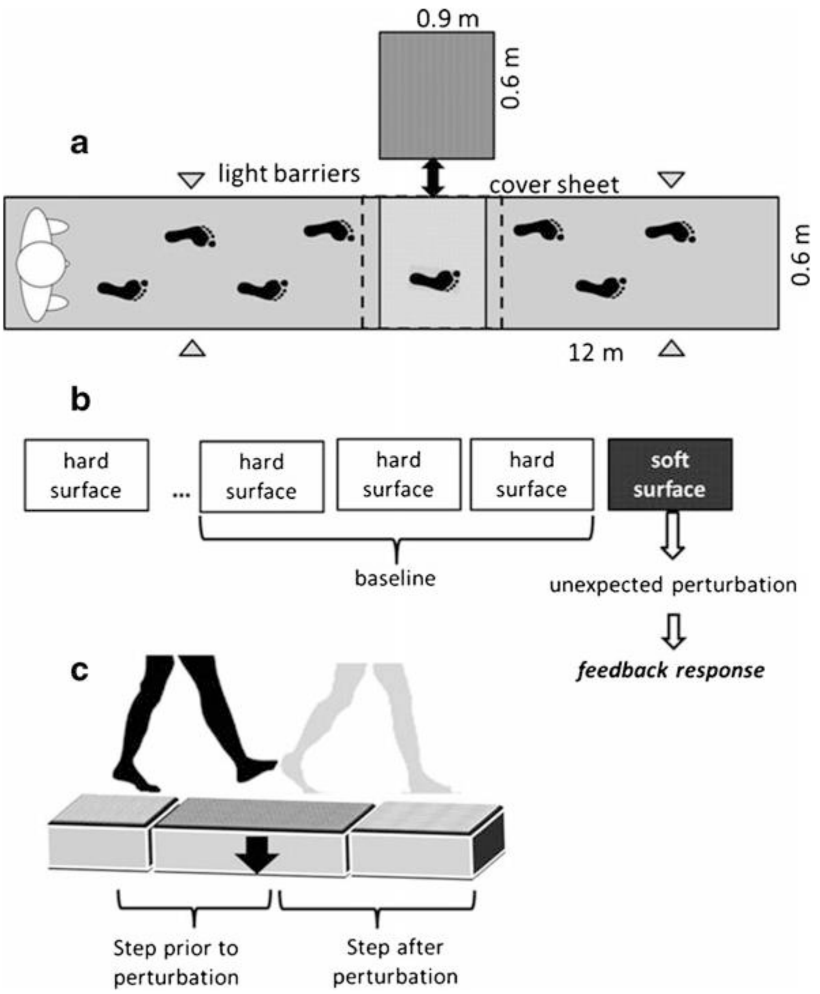

Figure 6-2: *Walkway (a) and gait protocol (b) for the test of the dynamic stability. The walkway (a) included one covered, exchangeable element, which allowed changing the surface condition from hard to soft without the knowledge of the participants. Three valid gait trials on the hard surface formed the baseline and were followed by one unexpected soft surface trial (b). This soft trial was performed to induce reactive behavior. Analysis was made for the step prior to the perturbation and the step after the perturbation (c).*

A meta-analysis identified 1.13 – 1.33 m/s as an average velocity for persons aged 60-79 years *(Bohannon & Andrews 2011)*. Therefore, we chose 1.3 m/s as gait velocity to pretend a moderate gait speed for the participants. The target velocity was controlled by light barriers and a stick which moved 1.3 m/s in front of the subjects along the gangway. The participants were informed that something in the walkway might change but they were not informed about the point of time and the type of the perturbation. After at least three valid gait trials on the hard surface (baseline) the participants experienced one unexpected soft surface trial. This trial was performed to detect feedback responses, because, up to this perturbation, the participants were used to the hard surface and had adjusted their behavior accordingly. Only one unexpected perturbation was analyzed due to the influence of expectations and predictive behavior on following trials *(Bierbaum et al. 2010, 2011; Marigold & Patla 2002)*.

6.3.4 Quantification of dynamic stability control

Kinematic data were recorded with the Vicon motion capture system (Model 624, Vicon, Oxford, UK) using 12 cameras operating at 120 Hz. The marker model contained 21 reflective markers (diameter 14 mm), which were fixed at specified positions. The marker trajectories were smoothed out using a Woltring filter routine *(Woltring 1986)* with a noise level of 10 mm^2. Segmental masses, the location of the segment centers of mass and the center of mass were calculated by a custom-written Matlab-model, based on the data reported by *Dempster et al. (1959).*

Since we wanted to analyze the stability state during single steps, we used the *"extrapolated center of mass"*-concept formulated by *Hof et al. (2005)* for the quantification of dynamic stability. This concept, based on the approach of *Pai & Patton (1997)*, is also applicable to disturbed movements. The margin of stability as a criterion for the state of postural stability of the human body was calculated as follows:

$MoS = U_{max} - X_{CM}$

MoS indicates the margin of stability in the anterior-posterior direction, U_{max} the anterior boundary of the base of support and X_{CM} the position of the extrapolated center of mass in the anterior-posterior direction ($X\_CM = P\_xCM + (V\_xCM/\sqrt{(g/l)})$). P_{xCM} is the horizontal (antero-posterior) component of the projection of the center of mass (CM) to the ground, V_{xCM} is the horizontal CM-velocity and the term $\sqrt{g/l}$ presents the eigenfrequency of a system of length l (inverted pendulum model). Margin of stability was analyzed only in the antero-posterior direction since the extrapolated center of mass shifts primarily in the anterior direction after the perturbation due to the increased velocity of the CM. MoS as well as the other components of dynamic stability are presented as the average values from touchdown of one leg to touchdown of the other leg to get a more representative stability state during walking. $Mean_{dist}$ labels the phase from touchdown left to touchdown of the disturbed, right leg and $mean_{rec}$ labels the phase from touchdown of the disturbed leg to touchdown of the recovery (left) leg.

For the investigation of the muscle strength of knee extensors (QF) and ankle flexors (TS), the participants performed maximum voluntary isometric contractions (MVIC) on a dynamometer (Biodex-System3, Biodex Medical Systems Inc., Shirley, NY, USA). Based on the torque–angle relation curve, the subjects performed the MIVC's near the optimal joint angle (i.e. plantar flexion MVIC*: ankle angle: 85°, knee angle: 180°*; knee extension MVIC: *knee angle: 120°, hip angle: 110°*), pre and post measurement. Resultant joint moments have been calculated according to a previously described method *(Arampatzis et al. 2004, 2005)*. By the use of this method, the antagonistic moment and the displacement of the joint axis in relation to the dynamometer axis are taken into account. Because of technical problems, muscle strength values are preserved for 12 subjects of the ST-group, 11 (QF) and 12 (TS) subjects of the MT-group and not for the control group.

A two-factor repeated-measures ANOVA with time (Pre vs. Post) as inter-subject variables and group as between-subject factors was used to examine the intervention effects on the analyzed dynamic stability parameters. Anthropometric data was analyzed by a one-way ANOVA and Bonferroni post-hoc comparisons between the groups. The level of significance was set to $\alpha = 0.05$.

6.4 Results

Age, body height, and body mass showed no differences between the three groups at the pre measurement. Furthermore, no difference between pre and post-test was found for the body mass (table 6-1). The ST-group showed no statistically significant differences in the pre/post comparisons of the maximum ankle plantar flexion moment (TS) as well as maximum knee extension moment (QF) [QF: $F(1, 11) = 0.159$, $p = 0.698$, part.$\eta^2 = 0.014$; TS: $F(1, 11) = 4.564$, $p = 0.056$, part.$\eta^2 = 0.293$]. The MT-group, on the other hand, increased significantly [$F(1, 10) = 32.582$, $p < 0.001$, part.$\eta^2 = 0.765$] the maximum isometric knee extension moments after the intervention but not the maximum ankle plantar flexion moment [$F(1, 11) = 1.089$, $p = 0.319$, part.$\eta^2 = 0.090$; table 6-1].

6.4.1 Baseline (unperturbed trials)

The gait velocity of the baseline trials showed a statistical significant main effect of time [$F(1, 35) = 11.205$, $p = 0.002$, part.$\eta^2 = 0.243$; table 6-2]. A closer look reveals that only the CG showed a significant increased gait velocity in the post measurement [follow up $F(1, 9) = 11.205$, $p = 0.009$, part.$\eta^2 = 0.555$]. However, the slightly increased gait velocity (~2%) in the post measurement did not significantly affect ($p > 0.05$) the margin of stability and the components of dynamic stability in the pre/post comparison (table 6-2). Therefore, no significant differences between pre- and post-test are observable in the remaining stability parameters for all groups.

6.4.2 Perturbed trial – last step prior to the perturbation

In the time frame prior to the unexpected perturbation in the soft surface condition, the parameters of dynamic stability did not show any statistically significant differences between the pre- and post-measurements. This indicates a state of similar gait stability before and after the intervention for all groups in the unexpected trials (table 6-3).

Table 6-1: *Anthropometric data, maximal isometric ankle flexion and knee extension moment at pre and post measurement for the experimental groups (mean ± SD)*

Intervention	ST (n=14) Pre	Post	MT (n=14) Pre	Post	CG (n=10) Pre	Post
Age (years)	67.4 ± 2.7		68.6 ± 3.1		68.9 ± 3.2	
Body height (cm)	165 ± 7.0		163.5 ± 7.3		168.6 ± 9.2	
Body weight (kg)	70.7 ± 9.5	70.4 ± 9.4	73.7 ± 13.0	72.8 ± 12.2	76.9 ± 13.4	76.7 ± 12.9
Max. Ankle flexion moment (Nm/kg)	1.97 ± 0.45	2.15 ± 0.45	1.94 ± 0.33	1.99 ± 0.31		
Max. Knee extension moment (Nm/kg) [a]	2.4 ± 0.46	2.43 ± 0.39	2.19 ± 0.36[b]	2.47 ± 0.28[b]		

[a] statistically significant interaction between time point of measurement and intervention group [repeated measurements; $F(1, 21) = 6.234$, $p=0.021$, part.$\eta^2 = 0.229$]

[b] statistically significant difference between pre and post measurement [$F(1, 10) = 32.582$, $p < 0.001$, part.$\eta^2 = 0.765$]

Table 6-2: *Mean values of the stability parameters (mean ± SD) in the three examined groups before (pre) and after (post) the intervention in the last step before the exchangeable element (time frame between touchdown left and touchdown of the right leg,* ***baseline (unperturbed trials)***

Intervention	ST (n=14) Pre	Post	MT (n=14) Pre	Post	CG (n=10) Pre	Post
Margin of stability ($mean_{dist}$) [cm]	-22.9 ± 3.8	-22.2 ± 3.7	-21.0 ± 5.3	-21.6 ± 6.3	-22.8 ± 5.5	-24.7 ± 5.0
CoM-velocity ($mean_{dist}$) [m/s] [a]	1.29 ± 0.04	1.31 ± 0.04	1.28 ± 0.04	1.3 ± 0.05	1.29 ± 0.03[b]	1.32 ± 0.03[b]
extrapolated CoM ($mean_{dist}$) [cm]	110.3 ± 6.2	110.0 ± 7.2	107.4 ± 5.9	109.3 ± 8.7	112.6 ± 8.2	113.8 ± 8.2
Base of support ($mean_{dist}$) [cm]	87.5 ± 5.4	87.8 ± 6.0	86.5 ± 3.7	87.8 ± 4.3	89.8 ± 5.3	89.1 ± 5.7
projected CoM ($mean_{dist}$) [cm]	72.1 ± 5.0	71.3 ± 6.1	70.3 ± 4.8	71.6 ± 7.1	74.3 ± 6.2	74.4 ± 6.3
term ($mean_{dist}$) [s^{-1}]	3.39 ± 0.09	3.39 ± 0.09	3.44 ± 0.09	3.45 ± 0.09	3.37 ± 0.15	3.37 ± 0.14

[a] statistically significant time effect time [$F(1, 35) = 11.205$, $p = 0.002$, part.$\eta^2 = 0.243$]
[b] statistically significant differences between pre and post measurement [follow up; $F(1, 9) = 11.205$, $p = 0.009$, part.$\eta^2 = 0.555$]

Table 6-3: *Mean values of the stability parameters (mean ± SD) in the three examined groups before (pre) and after (post) the intervention in the last step before the perturbation (time frame between touchdown left and touchdown of the right, disturbed leg),* ***soft surface trial***

Intervention	ST (n=14)		MT (n=14)		CG (n=10)	
	Pre	Post	Pre	Post	Pre	Post
Margin of stability ($mean_{dist}$) [cm]	-23.2 ± 5.3	-23.4 ± 4.3	-21.1 ± 5.8	-22.5 ± 6.7	-23.8 ± 4.6	-25.1 ± 4.8
CoM-velocity ($mean_{dist}$) [m/s]	1.29 ± 0.06	1.31 ± 0.07	1.28 ± 0.05	1.3 ± 0.07	1.3 ± 0.03	1.33 ± 0.04
extrapolated CoM ($mean_{dist}$) [cm]	111.3 ± 6.7	110.3 ± 7.4	106.8 ± 6.7	109.7 ± 9.3	114.2 ± 6.8	115.1 ± 8.4
Base of support ($mean_{dist}$) [cm]	88.1 ± 5.2	86.9 ± 6.5	85.7 ± 5.9	87.1 ± 4.5	90.4 ± 6.2	90.0 ± 6.3
projected CoM ($mean_{dist}$) [cm]	73.1 ± 5.4	71.8 ± 6.0	69.7 ± 5.8	72.1 ± 7.0	75.6 ± 5.4	75.6 ± 6.8
term ($mean_{dist}$) [s^{-1}]	3.38 ± 0.09	3.39 ± 0.08	3.45 ± 0.09	3.46 ± 0.09	3.37 ± 0.15	3.36 ± 0.14

Table 6-4: *Mean values of the stability parameters (mean ± SD) in the three examined groups before (pre) and after (post) the intervention in the first step after the unexpected perturbation (time frame between touchdown right, disturbed and touchdown left, recovery leg),* ***soft surface trial***

Intervention	ST (n=14) Pre	Post	MT (n=14) Pre	Post	CG (n=10) Pre	Post
CoM-velocity ($mean_{rec}$) [m/s]	1.42 ± 0.06	1.44 ± 0.08	1.39 ± 0.08	1.40 ± 0.05	1.40 ± 0.06	1.42 ± 0.06
Extrapolated CoM ($mean_{rec}$) [cm]	121.2 ± 7.8	122.3 ± 8.7	117.4 ± 6.7	118.5 ± 7.3	123.0 ± 10.0	123.6 ± 11.9
Base of support ($mean_{rec}$) [cm] [a]	90.9 ± 6.6[b]	98.2 ± 8.5[b]	91.4 ± 6.2	97.9 ± 12.7	97.9 ± 9.9	99.8 ± 9.3
Projected CoM ($mean_{rec}$) [cm]	77.5 ± 5.7	78.2 ± 7.0	75.3 ± 4.4	76.0 ± 5.6	79.8 ± 7.6	79.8 ± 8.8
Term ($mean_{rec}$) [s^{-1}]	3.26 ± 0.08	3.26 ± 0.08	3.3 ± 0.09	3.3 ± 0.08	3.25 ± 0.14	3.24 ± 0.13

[a] statistically significant time effect, [$F(1, 35) = 10.043$, $p = 0.003$, part.$\eta^2 = 0.223$]
[b] statistically significant differences between pre and post measurement [follow up; $F(1, 13) = 15.716$, $p = 0.002$, part.$\eta^2 = 0.547$]

6.4.3 Perturbed trial – first step after the perturbation

The repeated-measurements ANOVA showed a significant main effect of time [$F(1, 35) = 8.046$, $p = 0.008$, part.$\eta^2 = 0.187$] on the margin of stability at the step after the disturbance. A follow-up comparison revealed a significant increase in the margin of stability after the 14 weeks of training only in the ST-group [$F(1, 13) = 12.125$, $p = 0.004$, part.$\eta^2 = 0.483$; *fig. 6-3*].

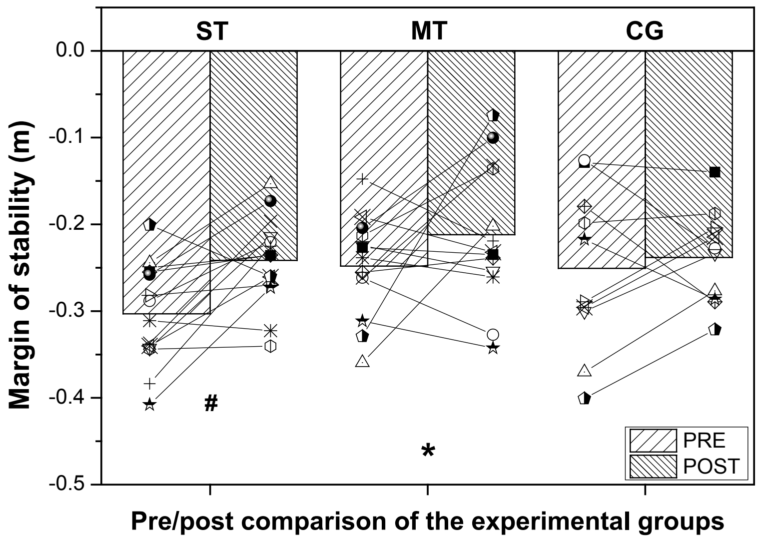

Figure 6-3: *Individual and mean values of margin of stability before and after the intervention at the step after the unexpected perturbation for the stability training group (ST), mixed training group (MT) and control group (CG).*
Asterisk, *statistically significant time effect, $F(1, 35) = 8.046$, $p=0.008$, part.$\eta^2=0.187$);* Number sign, *statistically significant differences between pre and post measurement [follow up; $F(1, 13) = 12.125$, $p = 0.004$, part.$\eta^2 = 0.483$]*

The increase in the margin of stability in the $mean_{rec}$-time frame after the intervention can be explained by a significant increase in the post-intervention base of support at the step after the unexpected perturbation in comparison to the pre-measurements [main effect of time; $F(1, 35) = 10.043$, $p = 0.003$, part.$\eta^2 = 0.223$]. The base of support was significantly increased from pre- to post-measurement in the ST-group [$F(1, 13) = 15.716$, $p = 0.002$, part.$\eta^2 = 0.547$] and showed a trend towards an increase in the MT-group [$F(1, 13) = 3.251$, $p = 0.095$, part.$\eta^2 = 0.200$; table 6-4]. The control group did not show any differences in the base of support before and after the 14 weeks [$F(1, 9) = 0.593$, $p = 0.461$, part.$\eta^2 = 0.062$]. All other parameters remained unaltered for all groups for the time course between touchdown of the disturbed leg and touchdown of the recovery leg.

6.5 Discussion

After 14 weeks of exercising the mechanisms of dynamic stability, both experimental groups achieved an average improvement of 18% in their stability performance after an unexpected gait perturbation. This supports our first hypothesis. The reason for the improvement in the stability state after the perturbation in the post-experiment was a significant increase of the base of support, indicating the importance of this mechanism regarding fall prevention. However, the mixed-exercise intervention group (i.e. training of mechanisms of dynamic stability and muscle strength) did not show clear improvements in the stability state compared to the pre-intervention values although the base of support after the unexpected perturbation showed a tendency for higher values in the post-condition. Therefore, the second hypothesis had to be rejected.

The rapid generation of high moments is an important factor for successful balance recovery after perturbations *(Pijnappels et al. 2005; Hsiao-Wecksler and Robinovitch 2007; Karamanidis et al. 2008; Mademli et al. 2008),* and therefore we expected the MT-group to achieve higher improvements in the dynamic stability in comparison to the ST-group. The adequate generation of the hip moment seems to be a major contributor to successful balance recovery *(Arampatzis et al. 2011).* Therefore, since the rectus femoris muscle as a knee extensor partially contributes to the hip moment, we assumed that the observed strength gain in the knee extensors via strength training (~13%) should possibly be reflected in an increased generation of hip moment. However, this increase in the knee extensor moment did not further increase the post-intervention recovery performance of the MT-group. Earlier studies reported a contribution of 30-40% of muscle strength to the capacity to recover balance with a single step after a forward fall *(Wojcik et al. 2001; Grabiner et al. 2005; Karamanidis et al. 2008)*. Therefore, we suggest that the increase of 13% of the knee extensor strength in our study is probably too low to achieve an improvement in the stability performance. A certain strength level is necessary to handle perturbations, but maximum leg strength appears not to be the critical factor. The main reason for the enhanced dynamic stability in simulated forward falls after an intervention period, for instance, was not the maximum joint moment but rather the ability to create a joint moment in an appropriate temporal framework *(Arampatzis et al. 2011)*.

Regarding the expected difference between the two training groups, the "less-than-expected"-development of the MT-group in comparison to the ST-group may be a consequence of the increased amount of training of the mechanisms of stability (1.5h vs. 45min.) for the ST-group. Beyond a certain strength level, seated-position strength training seems not to increase the performance in functional tasks such as the behavior after unexpected perturbations *(Sherrington et al. 2008)*. This means that alternative seated strength training for the lower extremities compensates not for the advantage of training the mechanisms of dynamic stability. However, as reduced muscle strength is an important risk factor for falls *(Moreland et al. 2004)*, strength training for the lower extremities still remains an important factor for individuals with muscle weakness *(Chandler et al. 1998; Latham et al. 2004)*.

In the ST-group, we found clear improvements in stability performance (~20%) during the unexpected gait perturbation after the intervention, but without any changes in muscle strength. This finding is in accordance with our earlier results during simulated forward falls *(Arampatzis et al. 2011)*, indicating that exercising the mechanisms of dynamic stability may lead to upgraded neuromuscular coordination and may be sufficient to convey successful strategies for regaining balance after perturbations. Improved capacity of maximal and explosive force production, as well as better neuronal activation, were already found in earlier sensorimotor training studies *(Granacher et al. 2007; Schubert et al. 2008)* and was explained by increased intermuscular coordination. Further, the repeated application of the mechanisms of dynamic stability seems to facilitate the adequate use of those mechanisms even in unfamiliar situations in order to regain balance. This improvement may be ascribed to a shift from prefrontal activity to a subcortical circuit, involving the cerebellum and basal ganglia, accompanied by increased automatic performance *(Floyer-Lea & Matthews 2004; Puttemans et al. 2005; Taube et al. 2007)*.

Several exercise interventions have so far shown to be effective in reducing the risk of falls in older adults by means of balance, coordination, or strength training *(for review see Gillespie et al. 2010).* Recently, especially perturbation-based training programs requiring reactive behavior show improvements in balance recovery *(Granacher et al. 2011)*. This is attributed to training specificity. However, our study was performed with the purpose of exercising specific mechanisms of dynamic stability, which are supposed to be fundamental to the regulation of postural stability. The participants did not exercise exact performance tasks, which were then measured, but were trained on exercises focusing on the underlying mechanisms of postural stability. Therefore, the intervention led to an improved application of the mechanisms of dynamic stability after a perturbation during walking, especially to an improved increase of the base of support. The performed perturbation during the measurements differed to the exercises applied in the training, indicating that the application of these mechanisms was transferable to a different situation.

However, we suggest that the imposed perturbation was relatively small because of the chosen gait velocity and the properties of the soft surface; higher effects may potentially be detected by applying more demanding tasks. Further, the demands of the ankle plantar flexion exercises for the MT group were possibly too low for a significant improvement of strength. Only one machine was available for the training of the ankle plantar flexors and therefore the participants had to exercise additionally with their own body weight which rather increased strength endurance. Ankle plantar flexion moment, however, is important during gait perturbations *(Pijnappels et al. 2004, 2005)* and an increase in this maximum moment could potentially support an increase in the base of support. The intervention groups showed no difference in their post values in both muscle groups (which may be caused by different initial strength conditions), but this does not affect the relevance of the performed training of the mechanisms of dynamic stability.

In conclusion, exercising the two mechanisms of dynamic stability *(increase of base of support & counter-rotating segments around the center of mass)* led to a better application of these mechanisms after an unexpected perturbation during gait. Significantly better postural stability after the intervention, though, was found only in the stability-training group; the training-group, which additionally exercised leg strength, did not show any further improvements. Therefore, we suggest that repeatedly exercising the mechanisms of dynamic stability contributes adequately to the accomplishment of improved dynamic stability, but that the performed additional strength training for healthy elderly individuals shows no further effect on the ability to recover balance after unexpected perturbations during gait.

Conflict of interest statement

The authors declare that they have no conflicts of interest.

6.6 References

Arampatzis, A., Karamanidis, K., and Mademli, L. (2008). Deficits in the way to achieve balance related to mechanisms of dynamic stability control in the elderly. J Biomech *41*, 1754–1761.

Arampatzis, A., Karamanidis, K., De Monte, G., Stafilidis, S., Morey-Klapsing, G., and Brüggemann, G.-P. (2004). Differences between measured and resultant joint moments during voluntary and artificially elicited isometric knee extension contractions. Clin Biomech (Bristol, Avon) *19*, 277–283.

Arampatzis, A., Morey-Klapsing, G., Karamanidis, K., DeMonte, G., Stafilidis, S., and Brüggemann, G.-P. (2005). Differences between measured and resultant joint moments during isometric contractions at the ankle joint. J Biomech *38*, 885–892.

Arampatzis, A., Peper, A., and Bierbaum, S. (2011). Exercise of mechanisms for dynamic stability control increases stability performance in the elderly. J Biomech *44*, 52–58.

Bhatt, T., Wening, J.D., and Pai, Y.-C. (2006). Adaptive control of gait stability in reducing slip-related backward loss of balance. Exp Brain Res *170*, 61–73.

Bierbaum, S., Peper, A., Karamanidis, K., and Arampatzis, A. (2010). Adaptational responses in dynamic stability during disturbed walking in the elderly. J Biomech *43*, 2362–2368.

Bierbaum, S., Peper, A., Karamanidis, K., and Arampatzis, A. (2011). Adaptive feedback potential in dynamic stability during disturbed walking in the elderly. J Biomech *44*, 1921–1926.

Blake, A.J., Morgan, K., Bendall, M.J., Dallosso, H., Ebrahim, S.B., Arie, T.H., Fentem, P.H., and Bassey, E.J. (1988). Falls by elderly people at home: prevalence and associated factors. Age Ageing *17*, 365–372.

Bohannon, R.W., and Andrews, A.W. (2011). Normal walking speed: a descriptive meta-analysis. Physiotherapy *97*, 182–189.

Chandler, J.M., Duncan, P.W., Kochersberger, G., and Studenski, S. (1998). Is lower extremity strength gain associated with improvement in physical performance and disability in frail, community-dwelling elders? Arch Phys Med Rehabil *79*, 24–30.

Chodzko-Zajko, W.J., Proctor, D.N., Fiatarone Singh, M.A., Minson, C.T., Nigg, C.R., Salem, G.J., and Skinner, J.S. (2009). American College of Sports Medicine position stand. Exercise and physical activity for older adults. Med Sci Sports Exerc *41*, 1510–1530.

Dempster, W.T., GABEL, W.C., and FELTS, W.J. (1959). The anthropometry of the manual work space for the seated subject. Am J Phys Anthropol *17*, 289–317.

Etman, A., Wijlhuizen, G.J., van Heuvelen, M.J.G., Chorus, A., and Hopman-Rock, M. (2012). Falls incidence underestimates the risk of fall-related injuries in older age groups: a comparison with the FARE (Falls risk by Exposure). Age Ageing *41*, 190–195.

Feder, G., Cryer, C., Donovan, S., and Carter, Y. (2000). Guidelines for the prevention of falls in people over 65. BMJ *321*, 1007–1011.

Floyer-Lea, A., and Matthews, P.M. (2004). Changing brain networks for visuomotor control with increased movement automaticity. J Neurophysiol *92*, 2405–2412.

Gillespie, L.D., Robertson, M.C., Gillespie, W.J., Lamb, S.E., Gates, S., Cumming, R.G., and Rowe, B.H. (2009). Interventions for preventing falls in older people living in the community. Cochrane Database Syst Rev CD007146.

Grabiner, M.D., Owings, T.M., and Pavol, M.J. (2005). Lower extremity strength plays only a small role in determining the maximum recoverable lean angle in older adults. J Gerontol A Biol Sci Med Sci *60*, 1447–1450.

Granacher, U., Gruber, M., and Gollhofer, A. (2009). Auswirkungen von sensomotorischem Training auf die posturale Kontrolle älterer Männer. Deutsche Zeitschrift für Sportmedizin *60*, 387–393.

Granacher, U., Muehlbauer, T., Zahner, L., Gollhofer, A., and Kressig, R.W. (2011). Comparison of traditional and recent approaches in the promotion of balance and strength in older adults. Sports Med *41*, 377–400.

Heiden, T.L., Sanderson, D.J., Inglis, J.T., and Siegmund, G.P. (2006). Adaptations to normal human gait on potentially slippery surfaces: the effects of awareness and prior slip experience. Gait Posture *24*, 237–246.

Heinrich, S., Weigelt, I., Rapp, K., Becker, C., Rissmann, U., and König, H.-H. (2012). Sturz- und Frakturprävention auf der Grundlage des Nationalen Expertenstandards Sturzprophylaxe. Zeitschrift für Gerontologie und Geriatrie *45*, 128–137.

Hof, A.L., Gazendam, M.G.J., and Sinke, W.E. (2005). The condition for dynamic stability. J Biomech *38*, 1–8.

Horak, F.B. (2006). Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? Age Ageing *35*, ii7–ii11.

Hortobágyi, T., Zheng, D., Weidner, M., Lambert, N.J., Westbrook, S., and Houmard, J.A. (1995). The influence of aging on muscle strength and muscle fiber characteristics with special reference to eccentric strength. J Gerontol A Biol Sci Med Sci *50*, B399–406.

Hsiao-Wecksler, E.T., and Robinovitch, S.N. (2007). The effect of step length on young and elderly women’s ability to recover balance. Clin Biomech *22*, 574–580.

Hunter, G.R., McCarthy, J.P., and Bamman, M.M. (2004). Effects of resistance training on older adults. Sports Med. *34*, 329–348.

Karamanidis, K., and Arampatzis, A. (2007). Age-related degeneration in leg-extensor muscle-tendon units decreases recovery performance after a forward fall: compensation with running experience. Eur J Appl Physiol *99*, 73–85.

Karamanidis, K., Arampatzis, A., and Mademli, L. (2008). Age-related deficit in dynamic stability control after forward falls is affected by muscle strength and tendon stiffness. J Electromyogr Kinesiol *18*, 980–989.

Latham, N.K., Bennett, D.A., Stretton, C.M., and Anderson, C.S. (2004). Systematic review of progressive resistance strength training in older adults. J Gerontol A Biol Sci Med Sci *59*, 48–61.

Lord, S.R., Ward, J.A., Williams, P., and Anstey, K.J. (1993). An epidemiological study of falls in older community-dwelling women: the Randwick falls and fractures study. Aust J Public Health *17*, 240–245.

Mademli, L., Arampatzis, A., and Karamanidis, K. (2008). Dynamic stability control in forward falls: postural corrections after muscle fatigue in young and older adults. Eur J Appl Physiol *103*, 295–306.

Maki, B.E., and McIlroy, W.E. (1997). The role of limb movements in maintaining upright stance: the “change-in-support” strategy. Phys Ther *77*, 488–507.

Marigold, D.S., and Patla, A.E. (2002). Strategies for dynamic stability during locomotion on a slippery surface: effects of prior experience and knowledge. J Neurophysiol *88*, 339–353.

Moreland, J.D., Richardson, J.A., Goldsmith, C.H., and Clase, C.M. (2004). Muscle weakness and falls in older adults: a systematic review and meta-analysis. J Am Geriatr Soc *52*, 1121–1129.

Pai, Y.C., and Patton, J. (1997). Center of mass velocity-position predictions for balance control. J Biomech *30*, 347–354.

Pijnappels, M., Bobbert, M.F., and van Dieën, J.H. (2004). Contribution of the support limb in control of angular momentum after tripping. J Biomech *37*, 1811–1818.

Pijnappels, M., Bobbert, M.F., and Dieën, J.H. van (2005). Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. Gait & Posture *21*, 388–394.

Puttemans, V., Wenderoth, N., and Swinnen, S.P. (2005). Changes in brain activation during the acquisition of a multifrequency bimanual coordination task: from the cognitive stage to advanced levels of automaticity. J Neurosci *25*, 4270–4278.

Robinovitch, S.N., Heller, B., Lui, A., and Cortez, J. (2002). Effect of strength and speed of torque development on balance recovery with the ankle strategy. J Neurophysiol *88*, 613–620.

Rubenstein, L.Z. (2006). Falls in older people: epidemiology, risk factors and strategies for prevention. Age Ageing *35 Suppl 2*, ii37–ii41.

Schubert, M., Beck, S., Taube, W., Amtage, F., Faist, M., and Gruber, M. (2008). Balance training and ballistic strength training are associated with task-specific corticospinal adaptations. Eur J Neurosci *27*, 2007–2018.

Sherrington, C., Whitney, J.C., Lord, S.R., Herbert, R.D., Cumming, R.G., and Close, J.C.T. (2008). Effective Exercise for the Prevention of Falls: A Systematic Review and Meta-Analysis. J Am Geriatr Soc *56*, 2234–2243.

Shumway-Cook, A., and Woollacott, M.H. (2007). Motor Control - Translating Research into Clinical Practice (Philadelphia: Lippincott Williams & Wilkins).

Taube, W., Gruber, M., Beck, S., Faist, M., Gollhofer, A., and Schubert, M. (2007). Cortical and spinal adaptations induced by balance training: correlation between stance stability and corticospinal activation. Acta Physiol (Oxf) *189*, 347–358.

Thelen, D.G., Wojcik, L.A., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (1997). Age differences in using a rapid step to regain balance during a forward fall. J Gerontol A Biol Sci Med Sci *52*, M8–13.

Tinetti, M.E., Speechley, M., and Ginter, S.F. (1988). Risk factors for falls among elderly persons living in the community. N Engl J Med *319*, 1701–1707.

Tseng, S.-C., Stanhope, S.J., and Morton, S.M. (2009). Impaired reactive stepping adjustments in older adults. J Gerontol A Biol Sci Med Sci *64*, 807–815.

Wojcik, L.A., Thelen, D.G., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (2001). Age and gender differences in peak lower extremity joint torques and ranges of motion used during single-step balance recovery from a forward fall. J Biomech *34*, 67–73.

Woltring, H.J. (1986). A Fortran package for generalized, cross-validatory spline smoothing and differentiation. Adv Eng Software (1978) *8*, 104–113.

7. Main findings and conclusion

This thesis was guided by the aim to gain enhanced knowledge about the sensorimotor adaptability in old age. The performed studies showed that elderly individuals preserve predictive as well as reactive sensorimotor adaptability in consequence to perturbations during walking. This means that older adults are able to learn how to adjust their dynamic stability behavior in succession to predictable and unpredictable perturbations.

By the use of two different gait protocols, this thesis was able to investigate the predictive as well as reactive adaptability of older adults in comparison to young. After the experience of repeated perturbations, older individuals were able to show similar predictive adaptive improvements compared to young adults. The main adaptation factor for both age-groups in the recovery performance during the walking protocol with the predictable perturbations was the size of the base of support. The mechanism *"increase of the base of support"* was applied prior to the perturbation and furthermore in consequence to the current undergone perturbation to achieve an overall more stable body position. The increased step length in preparation for the perturbation was applied after initially using a more cautious strategy, namely a more posterior position of the center of mass in relation to the base of support. Increasing the base of support is thought to be an adequate mechanism prior to the perturbation induced in these studies as well as afterwards due to the main movement and acceleration of the center of mass in the walking direction in the performed gait protocol.

In response to repeated unpredictable perturbations, older adults showed to be able to adapt their reactive behavior according to the demands of the experienced perturbation. However, compared to young individuals, older individuals showed a less variable locomotor behavior after the induced unexpected perturbations. This rather conservative behavior led to disadvantages in the reactive adaptation with respect to dynamic stability for the elderly. In the course of several unexpected perturbations (without observable predictive adjustments), young adults increased the base of support of their recovery step within a shorter duration, indicating a rapid torque generation. Old adults, however, rather used their initial successful strategy and showed only slight adjustments in the size of their base of support and the horizontal velocity of their center of mass in the course of repeated unexpected perturbations. Those adjustments helped to cope better with the consequence of the perturbation but were not sufficient enough to reach similar effectiveness compared to the young.

When considering the objectives of a fall prevention training program for older adults, it seems important to achieve a transfer effect of performed exercises. Assuming that the mentioned mechanisms of dynamic stability are underlying mechanisms for the control of postural stability, exercising those mechanisms should provide a basis for the generalization of the required strategies. In consequence to the performed training of the mechanisms of dynamic stability it could be observed that older adults are able to show a better recovery performance after unpredictable perturbations during gait. This is an indication for a generalization of the application of the exercised mechanisms to various situations, because the training program did not include the perturbed movement which was analyzed in the measurement sessions. Overall, the successful transfer of the learned capability indicates a generalization of the performed mechanisms into various situations and furthermore indicates a preserved adaptability for the old population.

The less-than-expected improvement in the dynamic stability following a mixed training program emphasizes the importance of the specificity of training. Strength training exercises at machines seem to have no big influence on the balance capability for healthy old adults, presuming a certain strength level. Exercising the mechanisms of dynamic stability within the stability training program twice as much as within the mixed training program seems to generate a higher advantage for the dynamic stability than the combined training of stability and seated-position strength exercises. The performance of seated-position strength training in place of exercising the mechanisms of dynamic stability for half of the time compensates not for the missing challenges for the dynamic stability control, but rather increases muscle strength in non-functional contexts.

The finding of short- and long-term reactive adaptability to perturbations which require reactive behavior furthermore supports the assumption of supraspinal control even during those fast movements. Earlier studies assumed that the experience of perturbations may cause alterations in the cortical control and therefore in the response pattern after the experience of perturbations *(McIlroy & Maki 1995; Quintern et al. 1985)*. This thesis found a change in the response pattern, namely a better application of the mechanisms of dynamic stability, after the experience of several perturbations. The repeated handling of perturbations which demand different postural strategies may facilitate the selection and execution of an adequate strategy within an appropriate time frame, improving the reactive behavior of old participants to regain balance after unexpected perturbations. The experience of *similar* perturbations may help to adjust the parameters of the movement pattern according to the wished outcome. And experiencing *different* perturbations may help to select and apply the adequate strategy in this specific context. It is thought that motor practice results in a better representation of the relation between motor commands and movements *(see Shadmehr & Mussa-Ivaldi 1994; Wang et al. 2011)*. Therefore, it can be suggested that the CNS improves the representation of the position and motion of the center of mass relative to the base of support and is able to select and execute the most adequate strategy in consequence to the training program. The neural adaptation to balance training has shown to include a shift from an initial high activation of the motor cortex to an enhanced contribution of subcortical regions. This could imply that in the beginning of the learning process the motor cortex improves the internal representations and learns to better predict the adequate motor output for the specific situation. With learning, processing of perturbations and their predicted consequences shifts to subcortical regions and the application of mechanisms of dynamic stability gets more and more automatic.

7.1 Practical implications

With regard to practical implications and recommendations, this thesis points to the importance of perturbation-based exercises in training programs for the prevention of falls in the elderly. In general, the training program should rely on the principles of motor learning such as individualization, overload, adaptation-progression and variability. The performed exercises should challenge the application of the mechanisms of dynamic stability in different contexts and conditions. Exercises can be created and selected according to the criterion "inclusion of the mechanisms of dynamic stability". Those exercises then can be adapted in their intensity to different skill levels by changing the size of the base of support or the size or the direction of the perturbations. Gradually reducing the base of support (e.g., two-legged stand, semi-tandem stand, tandem stand, one-legged stand, standing on a beam) increases the difficulty of the posture. Additional perturbations for exercising the mechanisms "counter-rotating segments around the center of mass" may be, for example, the application of external force by pushing or by throwing or catching balls. Larger perturbations may lead to the application of the mechanism "increase of base of support". Exercises may also be performed during walking on even or uneven surface, on a balance beam or on moveable platforms.

A mixed training program possibly shows an advantage when increasing the percentage of training exercises which include the mechanisms of dynamic stability compared to the mixed training program applied in this thesis. Furthermore, changing the conditions of strength training could help to increase the positive effect. Seated-position strength training should be replaced by functional strength training or by an inclusion of strengthening exercises into exercises for the training of the mechanisms of dynamic stability.

The training sessions were combined of group and individualized exercises, built up as a circle of postural stabilization tasks. This combination has shown to be motivating and interesting for the participating adults. Subjects, who quitted the training program at an early stage mainly left because of health problems and not because they disliked the training program. Altogether, the compliance was quite high and the training exercises showed to be applicable in different stages of learning and with different strength levels.

7.2 Limitations

In this thesis, the dynamic stability was measured and analyzed according to the concept of the extrapolated center of mass *(Hof et al. 2005)*. This concept allows the quantification of dynamic stability at discrete time points and therefore the quantification of consequences of postural perturbations. However, the concept by *Hof et al. (2005)* and *Hof (2008),* based on the inverted pendulum model, simplifies some characteristics of the human system. This means that the calculated stability state of this concept may show some inaccuracies compared to the actual state of the more complex human system. One simplification, for example, is the disregard of the double-contact period during walking (double-contact time is zero). During this period, though, the center of mass can be accelerated or decelerated and this influences the following behavior. Additionally, this concept does not account for accelerations of the arm or leg during the swing phase. Large excursions of the center of mass in relation to the pendulum length may further increase the inaccuracy of the calculations. Another limitation of the applied approach is the disregard of the ground reaction forces and the instantaneous torques at the ankle, knee and hip at touchdown of the disturbed leg and touchdown of the recovery step. Analyses of kinetic behavior and muscle activation also may contribute to a better understanding of the recovery processes and the predictive and reactive adjustments. Yet, regarding the performed studies we may say that, despite the aforementioned constraints, the applied concept of the extrapolated center of mass is able to quantify dynamic stability and to represent adaptation processes with respect to dynamic stability. The limitations of the selected approach should not affect the main findings of this thesis because of a similar systematic error for all investigated subjects.

Additionally, the chosen perturbations for the measurement sessions showed to have some disadvantages. First, the fixed position of the exchangeable element made it necessary to control the walking pattern of the subjects. As this was not always possible, subjects sometimes did not step in the required place on the exchangeable element and the performed trial had to be excluded. Second, the nature of the perturbation allowed quite slow responses. The perturbation induced by the soft surface led to a forward acceleration of the center of mass beyond the relatively fixed foot of the support leg. Since the induced perturbation can be perceived early in the stance phase of the disturbed leg, the remaining time left for the recovery behavior (before colliding with the walkway) is rather long. Third, the induced perturbation depended on the body weight of the subjects – light subjects experienced a smaller perturbation compared to heavy subjects. The first point resulted in the exclusion of several subjects from further analysis – thus, the remaining subjects walked with a normal and regular gait pattern and did not change their step length in advance of the exchangeable element. The second point indeed resulted in the aforementioned characteristics, but it affected all participants in the same amount so that there should be no influence on the results of the study. Thirdly, the participants' weight should have no influence on the results because of similar body weight in each of the investigated groups.

7.3 New questions and future lines of research

The current thesis aimed to gain insight into the predictive and reactive adaptability of elderly individuals. With regard to future studies, there are some open questions and proposed lines of future research.

We now know that older adults are able to learn an improved and adequate recovery behavior by the use of predictive and reactive strategies. However, in the current thesis the performed perturbations were unpredictable with respect to the timing but not regarding the spatial properties. We tried to control possible predictive behavior by analyzing the stability state at touchdown of the disturbed leg. However, attentional and arousal modifications also may affect the outcome of the perturbation. Therefore, with respect to future studies, it is desirable to apply perturbations at different time points and positions to be able to analyze purely reactive responses.

It is assumed that training of the mechanisms of dynamic stability leads to a better application of those mechanisms in diverse situations. The enhanced experience of the center of mass behavior relative to the base of support is suggested to improve postural control by improved cortical control, shifted later in the learning process to subcortical regions. These suggestions could be proven by dual task regimens on the one hand or by transcranial magnetic and also neural stimulation on the other hand. Regarding the assumed link between executive function and motor tasks, future studies should investigate the interplay of the application of underlying mechanisms of dynamic stability and executive functions. Previous studies suggested a reduced risk of developing dementia when walking daily *(Marquis et al. 2002; Veghese et al. 2002)* and, conversely, an increased risk of developing dementia for subjects with gait abnormalities *(Yaffe et al. 2001; van Gelder et al. 2004).* Further, faller-specific response to dual tasking was correlated with tests of executive functions *(Springer et al. 2006)*. Therefore, a relationship between dynamic stability and executive function can be suggested – keeping in mind that this relationship surely is multi-factorial *(Yogev-Seligman et al. 2008)*. Thus, it would be interesting to see if there is a causal relationship between dynamic stability and executive function. In addition, from a psychological point of view, there is still an open question where and how fear of falling effects the selection and execution of the adequate recovery strategies.

Furthermore, using the approach by *Hof et al. (2005)*, it is possible to analyze the percentual contribution of the mechanisms "moving the center of pressure" or "increase of base of support" and "counter-rotation of segments around the center of mass". This would provide interesting information regarding the behavior of old and young subjects and in consequence to repeated perturbations.

Regarding the training of dynamic stability, further studies should investigate the retention effect of the application of the mechanisms of dynamic stability. This thesis only investigated the recovery behavior of elderly adults direct afterwards the training program. It still remains unknown, if older subjects are able to preserve the learned improved recovery behavior.

8. References

Andrews-Hanna, J.R., Snyder, A.Z., Vincent, J.L., Lustig, C., Head, D., Raichle, M.E., and Buckner, R.L. (2007). Disruption of large-scale brain systems in advanced aging. Neuron *56*, 924–935.

Arampatzis, A., Karamanidis, K., and Mademli, L. (2008). Deficits in the way to achieve balance related to mechanisms of dynamic stability control in the elderly. J Biomech *41*, 1754–1761.

Arampatzis, A., Karamanidis, K., De Monte, G., Stafilidis, S., Morey-Klapsing, G., and Brüggemann, G.-P. (2004). Differences between measured and resultant joint moments during voluntary and artificially elicited isometric knee extension contractions. Clin Biomech (Bristol, Avon) *19*, 277–283.

Arampatzis, A., Morey-Klapsing, G., Karamanidis, K., DeMonte, G., Stafilidis, S., and Brüggemann, G.-P. (2005). Differences between measured and resultant joint moments during isometric contractions at the ankle joint. J Biomech *38*, 885–892.

Arampatzis, A., Peper, A., and Bierbaum, S. (2011). Exercise of mechanisms for dynamic stability control increases stability performance in the elderly. J Biomech *44*, 52–58.

Arnsten, A.F.T., and Li, B.-M. (2005). Neurobiology of executive functions: catecholamine influences on prefrontal cortical functions. Biol Psychiatry *57*, 1377–1384.

Auyeung, T.W., Kwok, T., Lee, J., Leung, P.C., Leung, J., and Woo, J. (2008). Functional decline in cognitive impairment--the relationship between physical and cognitive function. Neuroepidemiology *31*, 167–173.

Baloh, R.W., Enrietto, J., Jacobson, K.M., and Lin, A. (2001). Age-related changes in vestibular function: a longitudinal study. Ann N Y Acad Sci *942*, 210–219.

Barak, Y., Wagenaar, R.C., and Holt, K.G. (2006). Gait characteristics of elderly people with a history of falls: a dynamic approach. Phys Ther *86*, 1501–1510.

Barrett, R.S., Mills, P.M., and Begg, R.K. (2010). A systematic review of the effect of ageing and falls history on minimum foot clearance characteristics during level walking. Gait Posture *32*, 429–435.

Bastian, A.J. (2006). Learning to predict the future: the cerebellum adapts feedforward movement control. Curr Opin Neurobiol *16*, 645–649.

Bastian, A.J. (2008). Understanding sensorimotor adaptation and learning for rehabilitation. Curr Opin Neurol *21*, 628–633.

Benjuya, N., Melzer, I., and Kaplanski, J. (2004). Aging-induced shifts from a reliance on sensory input to muscle cocontraction during balanced standing. J Gerontol A Biol Sci Med Sci *59*, 166–171.

Berg, W.P., Alessio, H.M., Mills, E.M., and Tong, C. (1997). Circumstances and consequences of falls in independent community-dwelling older adults. Age Ageing *26*, 261–268.

Bhatt, T., Espy, D., Yang, F., and Pai, Y.-C. (2011). Dynamic gait stability, clinical correlates, and prognosis of falls among community-dwelling older adults. Arch Phys Med Rehabil *92*, 799–805.

Bhatt, T., and Pai, Y.-C. (2008). Can observational training substitute motor training in preventing backward balance loss after an unexpected slip during walking? J Neurophysiol *99*, 843–852.

Bhatt, T., Wening, J.D., and Pai, Y.-C. (2005). Influence of gait speed on stability: recovery from anterior slips and compensatory stepping. Gait Posture *21*, 146–156.

Bhatt, T., Wening, J.D., and Pai, Y.-C. (2006). Adaptive control of gait stability in reducing slip-related backward loss of balance. Exp Brain Res *170*, 61–73.

Bierbaum, S., Peper, A., Karamanidis, K., and Arampatzis, A. (2010). Adaptational responses in dynamic stability during disturbed walking in the elderly. J Biomech *43*, 2362–2368.

Bierbaum, S., Peper, A., Karamanidis, K., and Arampatzis, A. (2011). Adaptive feedback potential in dynamic stability during disturbed walking in the elderly. J Biomech *44*, 1921–1926.

Bishop, N.A., Lu, T., and Yankner, B.A. (2010). Neural mechanisms of ageing and cognitive decline. Nature *464*, 529–535.

Blake, A.J., Morgan, K., Bendall, M.J., Dallosso, H., Ebrahim, S.B., Arie, T.H., Fentem, P.H., and Bassey, E.J. (1988). Falls by elderly people at home: prevalence and associated factors. Age Ageing *17*, 365–372.

Bo, J., and Seidler, R.D. (2009). Visuospatial working memory capacity predicts the organization of acquired explicit motor sequences. J Neurophysiol *101*, 3116–3125.

Bock, O. (2005). Components of sensorimotor adaptation in young and elderly subjects. Exp Brain Res *160*, 259–263.

Bock, O., and Schneider, S. (2001). Acquisition of a sensorimotor skill in younger and older adults. Acta Physiol Pharmacol Bulg *26*, 89–92.

Bock, O., and Schneider, S. (2002). Sensorimotor adaptation in young and elderly humans. Neurosci Biobehav Rev *26*, 761–767.

Bogle Thorbahn, L.D., and Newton, R.A. (1996). Use of the Berg Balance Test to predict falls in elderly persons. Phys Ther *76*, 576–583.

Bohannon, R.W., and Andrews, A.W. (2011). Normal walking speed: a descriptive meta-analysis. Physiotherapy *97*, 182–189.

Bohm, S., Mersmann, F., Bierbaum, S., Dietrich, R., and Arampatzis, A. (2012). Cognitive demand and predictive adaptational responses in dynamic stability control. J Biomech 45(14):2330-2336.

Bolton, C.F., Winkelmann, R.K., and Dyck, P.J. (1966). A quantitative study of Meissner's corpuscles in man. Neurology *16*, 1–9.

Boulgarides, L.K., McGinty, S.M., Willett, J.A., and Barnes, C.W. (2003). Use of clinical and impairment-based tests to predict falls by community-dwelling older adults. Phys Ther *83*, 328–339.

Boyle, P.A., Buchman, A.S., Wilson, R.S., Leurgans, S.E., and Bennett, D.A. (2009). Association of muscle strength with the risk of Alzheimer disease and the rate of cognitive decline in community-dwelling older persons. Arch Neurol *66*, 1339–1344.

Brach, J.S., Berlin, J.E., VanSwearingen, J.M., Newman, A.B., and Studenski, S.A. (2005). Too much or too little step width variability is associated with a fall history in older persons who walk at or near normal gait speed. J Neuroeng Rehabil *2*, 21.

Brown, L.A., Shumway-Cook, A., and Woollacott, M.H. (1999). Attentional demands and postural recovery: the effects of aging. J Gerontol A Biol Sci Med Sci *54*, M165–171.

Brown, W., Strong, M., and Snow, R. (1988). Methods for Estimating Numbers of Motor Units in Biceps-Brachialis Muscles and Losses of Motor Units with Aging. Muscle Nerve *11*, 423–432.

Bruijn, S.M., Bregman, D.J.J., Meijer, O.G., Beek, P.J., and van Dieën, J.H. (2011). The validity of stability measures: a modelling approach. J Biomech *44*, 2401–2408.

Bruijn, S.M., van Dieën, J.H., Meijer, O.G., and Beek, P.J. (2009). Statistical precision and sensitivity of measures of dynamic gait stability. J Neurosci Methods *178*, 327–333.

Bruijn, S.M., Ten Kate, W.R.T., Faber, G.S., Meijer, O.G., Beek, P.J., and van Dieën, J.H. (2010). Estimating dynamic gait stability using data from non-aligned inertial sensors. Ann Biomed Eng *38*, 2588–2593.

Buch, E.R., Young, S., and Contreras-Vidal, J.L. (2003). Visuomotor adaptation in normal aging. Learn Mem *10*, 55–63.

Bunday, K.L., Reynolds, R.F., Kaski, D., Rao, M., Salman, S., and Bronstein, A.M. (2006). The effect of trial number on the emergence of the “broken escalator” locomotor aftereffect. Exp Brain Res *174*, 270–278.

Burleigh, A., and Horak, F. (1996). Influence of instruction, prediction, and afferent sensory information on the postural organization of step initiation. J Neurophysiol *75*, 1619–1628.

Buzzi, U.H., and Ulrich, B.D. (2004). Dynamic stability of gait cycles as a function of speed and system constraints. Motor Control *8*, 241–254.

Camicioli, R., Panzer, V.P., and Kaye, J. (1997). Balance in the healthy elderly: posturography and clinical assessment. Arch Neurol *54*, 976–981.

Carty, C.P., Cronin, N.J., Lichtwark, G.A., Mills, P.M., and Barrett, R.S. (2012). Mechanisms of adaptation from a multiple to a single step recovery strategy following repeated exposure to forward loss of balance in older adults. PLoS ONE *7*, e33591.

Caterino, J.M., Karaman, R., Arora, V., Martin, J.L., and Hiestand, B.C. (2009). Comparison of balance assessment modalities in emergency department elders: a pilot cross-sectional observational study. BMC Emerg Med *9*, 19.

Chandler, J.M., Duncan, P.W., Kochersberger, G., and Studenski, S. (1998). Is lower extremity strength gain associated with improvement in physical performance and disability in frail, community-dwelling elders? Arch Phys Med Rehabil *79*, 24–30.

Chang, M.D., Shaikh, S., and Chau, T. (2009). Effect of treadmill walking on the stride interval dynamics of human gait. Gait Posture *30*, 431–435.

Chapman, G.J., and Hollands, M.A. (2007). Evidence that older adult fallers prioritise the planning of future stepping actions over the accurate execution of ongoing steps during complex locomotor tasks. Gait Posture *26*, 59–67.

Chen, H.C., Ashton-Miller, J.A., Alexander, N.B., and Schultz, A.B. (1991). Stepping over obstacles: gait patterns of healthy young and old adults. J Gerontol A Biol Sci Med Sci *46*, M196–203.

Chen, H.C., Ashton-Miller, J.A., Alexander, N.B., and Schultz, A.B. (1994). Effects of age and available response time on ability to step over an obstacle. J Gerontol A Biol Sci Med Sci *49*, M227–233.

Chen, H.C., Schultz, A.B., Ashton-Miller, J.A., Giordani, B., Alexander, N.B., and Guire, K.E. (1996). Stepping over obstacles: dividing attention impairs performance of old more than young adults. J Gerontol A Biol Sci Med Sci *51*, M116–122.

Chodzko-Zajko, W.J., Proctor, D.N., Fiatarone Singh, M.A., Minson, C.T., Nigg, C.R., Salem, G.J., and Skinner, J.S. (2009). American College of Sports Medicine position stand. Exercise and physical activity for older adults. Med Sci Sports Exerc *41*, 1510–1530.

Christensen, L.O., Petersen, N., Andersen, J.B., Sinkjaer, T., and Nielsen, J.B. (2000). Evidence for transcortical reflex pathways in the lower limb of man. Prog Neurobiol *62*, 251–272.

Criswell, D.S., Powers, S.K., Herb, R.A., and Dodd, S.L. (1997). Mechanism of specific force deficit in the senescent rat diaphragm. Respir Physiol *107*, 149–155.

da Cruz, D.T., Ribeiro, L.C., Vieira, M. de T., Teixeira, M.T.B., Bastos, R.R., and Leite, I.C.G. (2012). Prevalence of falls and associated factors in elderly individuals. Rev Saude Publica *46*, 138–146.

Cruz-Jentoft, A.J., Baeyens, J.P., Bauer, J.M., Boirie, Y., Cederholm, T., Landi, F., Martin, F.C., Michel, J.-P., Rolland, Y., Schneider, S.M., et al. (2010). Sarcopenia: European Consensus on Definition and Diagnosis Report of the European Working Group on Sarcopenia in Older People. Age Ageing *39*, 412–423.

Danion, F., Varraine, E., Bonnard, M., and Pailhous, J. (2003). Stride variability in human gait: the effect of stride frequency and stride length. Gait Posture *18*, 69–77.

D'Antona, G., Pellegrino, M.A., Adami, R., Rossi, R., Carlizzi, C.N., Canepari, M., Saltin, B., and Bottinelli, R. (2003). The effect of ageing and immobilization on structure and function of human skeletal muscle fibres. J Physiol (Lond.) *552*, 499–511.

Deandrea, S., Lucenteforte, E., Bravi, F., Foschi, R., La Vecchia, C., and Negri, E. (2010). Risk factors for falls in community-dwelling older people: a systematic review and meta-analysis. Epidemiology *21*, 658–668.

Delbaere, K., Close, J.C.T., Heim, J., Sachdev, P.S., Brodaty, H., Slavin, M.J., Kochan, N.A., and Lord, S.R. (2010). A multifactorial approach to understanding fall risk in older people. J Am Geriatr Soc *58*, 1679–1685.

Delbono, O., O'Rourke, K.S., and Ettinger, W.H. (1995). Excitation-calcium release uncoupling in aged single human skeletal muscle fibers. J Membr Biol *148*, 211–222.

Dempster, W.T., Gabel, W.C., and Felts, W.J. (1959). The anthropometry of the manual work space for the seated subject. Am J Phys Anthropol *17*, 289–317.

Dempster, W.T., Gaughran, G.R.L. (1967). Properties of body segments based on size and weight. Am J Anatomy *120*(1): 33-54.

Dennis, N.A., Howard, J.H., Jr, and Howard, D.V. (2003). Age deficits in learning sequences of spoken words. J Gerontol B Psychol Sci Soc Sci *58*, P224–227.

Der, G., and Deary, I.J. (2006). Age and sex differences in reaction time in adulthood: results from the United Kingdom Health and Lifestyle Survey. Psychol Aging *21*, 62–73.

DeVita, P., and Hortobagyi, T. (2000). Age causes a redistribution of joint torques and powers during gait. J Appl Physiol *88*, 1804–1811.

Dietz, V., Quintern, J., and Berger, W. (1984). Cerebral evoked potentials associated with the compensatory reactions following stance and gait perturbation. Neurosci Lett *50*, 181–186.

Dietz, V., Quintern, J., and Sillem, M. (1987). Stumbling reactions in man: significance of proprioceptive and pre-programmed mechanisms. J Physiol (Lond.) *386*, 149–163.

Dimitriou, M., and Edin, B.B. (2010). Human muscle spindles act as forward sensory models. Curr Biol *20*, 1763–1767.

Dingwell, J.B., and Cusumano, J.P. (2000). Nonlinear time series analysis of normal and pathological human walking. Chaos *10*, 848–863.

Dingwell, J.B., Cusumano, J.P., Cavanagh, P.R., and Sternad, D. (2001). Local dynamic stability versus kinematic variability of continuous overground and treadmill walking. J Biomech Eng *123*, 27–32.

Dingwell, J.B., and Kang, H.G. (2007). Differences between local and orbital dynamic stability during human walking. J Biomech Eng *129*, 586–593.

Doherty, T., Vandervoort, A., Taylor, A., and Brown, W. (1993). Effects of Motor Unit Losses on Strength in Older Men and Women. J Appl Physiol *74*, 868–874.

Dubost, V., Kressig, R.W., Gonthier, R., Herrmann, F.R., Aminian, K., Najafi, B., and Beauchet, O. (2006). Relationships between dual-task related changes in stride velocity and stride time variability in healthy older adults. Hum Mov Sci *25*, 372–382.

Earhart, G.M., Fletcher, W.A., Horak, F.B., Block, E.W., Weber, K.D., Suchowersky, O., and Melvill Jones, G. (2002a). Does the cerebellum play a role in podokinetic adaptation? Exp Brain Res *146*, 538–542.

Earhart, G.M., Jones, G.M., Horak, F.B., Block, E.W., Weber, K.D., and Fletcher, W.A. (2002b). Podokinetic after-rotation following unilateral and bilateral podokinetic stimulation. J Neurophysiol *87*, 1138–1141.

Earles, D.R., Koceja, D.M., and Shively, C.W. (2000). Environmental changes in soleus H-reflex excitability in young and elderly subjects. Int J Neurosci *105*, 1–13.

Eckner, J.T., Richardson, J.K., Kim, H., Lipps, D.B., and Ashton-Miller, J.A. (2012). A novel clinical test of recognition reaction time in healthy adults. Psychol Assess *24*, 249–254.

Eng, J.J., Winter, D.A., and Patla, A.E. (1994). Strategies for recovery from a trip in early and late swing during human walking. Exp Brain Res *102*, 339–349.

England, S.A., and Granata, K.P. (2007). The influence of gait speed on local dynamic stability of walking. Gait Posture *25*, 172–178.

Erni, T., and Dietz, V. (2001). Obstacle avoidance during human walking: learning rate and cross-modal transfer. J Physiol (Lond.) *534*, 303–312.

Etman, A., Wijlhuizen, G.J., van Heuvelen, M.J.G., Chorus, A., and Hopman-Rock, M. (2012). Falls incidence underestimates the risk of fall-related injuries in older age groups: a comparison with the FARE (Falls risk by Exposure). Age Ageing *41*, 190–195.

Feder, G., Cryer, C., Donovan, S., and Carter, Y. (2000). Guidelines for the prevention of falls in people over 65. BMJ *321*, 1007–1011.

Fernández-Ruiz, J., and Díaz, R. (1999). Prism adaptation and aftereffect: specifying the properties of a procedural memory system. Learn Mem *6*, 47–53.

Fernández-Ruiz, J., Hall, C., Vergara, P., and Díiaz, R. (2000). Prism adaptation in normal aging: slower adaptation rate and larger aftereffect. Brain Res Cogn Brain Res *9*, 223–226.

Floyer-Lea, A., and Matthews, P.M. (2004). Changing brain networks for visuomotor control with increased movement automaticity. J Neurophysiol *92*, 2405–2412.

Forner-Cordero, A. (2003). Human gait, stumble and... fall? Mechanical limitations of the recovery from a stumble (Twente: Universiteit Twente, Dissertation).

Fozard, J.L., Vercryssen, M., Reynolds, S.L., Hancock, P.A., and Quilter, R.E. (1994). Age differences and changes in reaction time: the Baltimore Longitudinal Study of Aging. J Gerontol A Biol Sci Med Sci *49*, P179–189.

Frontera, W.R., Hughes, V.A., Lutz, K.J., and Evans, W.J. (1991). A cross-sectional study of muscle strength and mass in 45- to 78-yr-old men and women. J Appl Physiol *71*, 644–650.

Frontera, W.R., Suh, D., Krivickas, L.S., Hughes, V.A., Goldstein, R., and Roubenoff, R. (2000). Skeletal muscle fiber quality in older men and women. Am J Physiol, Cell Physiol *279*, C611–618.

Galna, B., Peters, A., Murphy, A.T., and Morris, M.E. (2009). Obstacle crossing deficits in older adults: a systematic review. Gait Posture *30*, 270–275.

van Gelder, B.M., Tijhuis, M.A.R., Kalmijn, S., Giampaoli, S., Nissinen, A., and Kromhout, D. (2004). Physical activity in relation to cognitive decline in elderly men: the FINE Study. Neurology *63*, 2316–2321.

Gibbs, J., Hughes, S., Dunlop, D., Singer, R., and Chang, R.W. (1996). Predictors of change in walking velocity in older adults. J Am Geriatr Soc *44*, 126–132.

Gillespie, L.D., Robertson, M.C., Gillespie, W.J., Lamb, S.E., Gates, S., Cumming, R.G., and Rowe, B.H. (2009). Interventions for preventing falls in older people living in the community. Cochrane Database Syst Rev CD007146.

Goble, D.J., Coxon, J.P., Wenderoth, N., Van Impe, A., and Swinnen, S.P. (2009). Proprioceptive sensibility in the elderly: degeneration, functional consequences and plastic-adaptive processes. Neurosci Biobehav Rev *33*, 271–278.

Grabiner, M.D., Owings, T.M., and Pavol, M.J. (2005). Lower extremity strength plays only a small role in determining the maximum recoverable lean angle in older adults. J Gerontol A Biol Sci Med Sci *60*, 1447–1450.

Granacher, U., and Gollhofer, A. (2005). Auswirkungen des Alterns auf die Schnellkraftfähigkeit und das Reflexverhalten - ein Übersichtsbeitrag. Deutsche Zeitschrift für Sportmedizin *56*, 68–72.

Granacher, U., Gollhofer, A., and Strass, D. (2006). Training induced adaptations in characteristics of postural reflexes in elderly men. Gait Posture *24*, 459–466.

Granacher, U., Gruber, M., and Gollhofer, A. (2009). Auswirkungen von sensomotorischem Training auf die posturale Kontrolle älterer Männer. Deutsche Zeitschrift für Sportmedizin *60*, 387–393.

Granacher, U., Muehlbauer, T., Zahner, L., Gollhofer, A., and Kressig, R.W. (2011). Comparison of traditional and recent approaches in the promotion of balance and strength in older adults. Sports Med *41*, 377–400.

Granata, K.P., and Lockhart, T.E. (2008). Dynamic stability differences in fall-prone and healthy adults. J Electromyogr Kinesiol *18*, 172–178.

Gruber, M., Taube, W., Gollhofer, A., Beck, S., Amtage, F., and Schubert, M. (2007). Training-specific adaptations of H- and stretch reflexes in human soleus muscle. J Mot Behav *39*, 68–78.

Gunning-Dixon, F.M., Brickman, A.M., Cheng, J.C., and Alexopoulos, G.S. (2009). Aging of cerebral white matter: a review of MRI findings. Int J Geriatr Psychiatry *24*, 109–117.

Gunning-Dixon, F.M., and Raz, N. (2003). Neuroanatomical correlates of selected executive functions in middle-aged and older adults: a prospective MRI study. Neuropsychologia *41*, 1929–1941.

Haddad, J.M., Gagnon, J.L., Hasson, C.J., Van Emmerik, R.E.A., and Hamill, J. (2006). Evaluation of time-to-contact measures for assessing postural stability. J Appl Biomech *22*, 155–161.

Hamacher, D., Singh, N.B., Van Dieën, J.H., Heller, M.O., and Taylor, W.R. (2011). Kinematic measures for assessing gait stability in elderly individuals: a systematic review. J R Soc Interface *8*, 1682–1698.

Hasson, C.J., Van Emmerik, R.E.A., and Caldwell, G.E. (2008). Predicting dynamic postural instability using center of mass time-to-contact information. J Biomech *41*, 2121–2129.

Hausdorff, J.M. (2007). Gait dynamics, fractals and falls: finding meaning in the stride-to-stride fluctuations of human walking. Hum Mov Sci *26*, 555–589.

Hausdorff, J.M., Edelberg, H.K., Cudkowicz, M.E., Singh, M.A., and Wei, J.Y. (1997). The relationship between gait changes and falls. J Am Geriatr Soc *45*, 1406.

Hausdorff, J.M., Peng, C.K., Ladin, Z., Wei, J.Y., and Goldberger, A.L. (1995). Is walking a random walk? Evidence for long-range correlations in stride interval of human gait. J Appl Physiol *78*, 349–358.

Hay, L., Bard, C., Fleury, M., and Teasdale, N. (1996). Availability of visual and proprioceptive afferent messages and postural control in elderly adults. Experimental Brain Research *108*, 129–139.

van Hedel, H., Biedermann, M., Erni, T., and Dietz, V. (2002). Obstacle avoidance during human walking: transfer of motor skill from one leg to the other. J Physiology *543*, 709–727.

Heiden, T.L., Sanderson, D.J., Inglis, J.T., and Siegmund, G.P. (2006). Adaptations to normal human gait on potentially slippery surfaces: the effects of awareness and prior slip experience. Gait Posture *24*, 237–246.

Heinrich, S., Rapp, K., Rissmann, U., Becker, C., and König, H.-H. (2010). Cost of falls in old age: a systematic review. Osteoporos Int *21*, 891–902.

Heinrich, S., Weigelt, I., Rapp, K., Becker, C., Rissmann, U., and König, H.-H. (2012). Sturz- und Frakturprävention auf der Grundlage des Nationalen Expertenstandards Sturzprophylaxe. Zeitschrift für Gerontologie und Geriatrie *45*, 128–137.

Herdman, S.J., Blatt, P., Schubert, M.C., and Tusa, R.J. (2000). Falls in patients with vestibular deficits. Am J Otol *21*, 847–851.

Herman, T., Giladi, N., Gurevich, T., and Hausdorff, J.M. (2005). Gait instability and fractal dynamics of older adults with a "cautious" gait: why do certain older adults walk fearfully? Gait Posture *21*, 178–185.

Hof, A.L. (2005). Comparison of three methods to estimate the center of mass during balance assessment. J Biomech *38*, 2134–2135.

Hof, A.L. (2007). The equations of motion for a standing human reveal three mechanisms for balance. J Biomech *40*, 451–457.

Hof, A.L. (2008). The "extrapolated center of mass" concept suggests a simple control of balance in walking. Hum Mov Sci *27*, 112–125.

Hof, A.L., Gazendam, M.G.J., and Sinke, W.E. (2005). The condition for dynamic stability. J Biomech *38*, 1–8.

Hollman, J.H., Kovash, F.M., Kubik, J.J., and Linbo, R.A. (2007). Age-related differences in spatiotemporal markers of gait stability during dual task walking. Gait Posture 26, 113–119.

Holt, K.J., Jeng, S.F., RR, R.R., and Hamill, J. (1995). Energetic Cost and Stability During Human Walking at the Preferred Stride Velocity. J Mot Behav *27*, 164–178.

Horak, F.B. (2006). Postural orientation and equilibrium: what do we need to know about neural control of balance to prevent falls? Age Ageing *35*, ii7–ii11.

Horak, F.B., Diener, H.C., and Nashner, L.M. (1989a). Influence of central set on human postural responses. J Neurophysiol *62*, 841–853.

Horak, F.B., Henry, S.M., and Shumway-Cook, A. (1997). Postural perturbations: new insights for treatment of balance disorders. Phys Ther *77*, 517–533.

Horak, F.B., and Nashner, L.M. (1986). Central programming of postural movements: adaptation to altered support-surface configurations. J Neurophysiol *55*, 1369–1381.

Horak, F.B., Shupert, C.L., and Mirka, A. (1989b). Components of postural dyscontrol in the elderly: a review. Neurobiol Aging *10*, 727–738.

Hortobágyi, T., Zheng, D., Weidner, M., Lambert, N.J., Westbrook, S., and Houmard, J.A. (1995). The influence of aging on muscle strength and muscle fiber characteristics with special reference to eccentric strength. J Gerontol A Biol Sci Med Sci *50*, B399–406.

Howard, D.V., and Howard, J.H., Jr (1992). Adult age differences in the rate of learning serial patterns: evidence from direct and indirect tests. Psychol Aging *7*, 232–241.

Howard, D.V., Howard, J.H., Jr, Japikse, K., DiYanni, C., Thompson, A., and Somberg, R. (2004). Implicit sequence learning: effects of level of structure, adult age, and extended practice. Psychol Aging *19*, 79–92.

Howe, T.E., Rochester, L., Neil, F., Skelton, D.A., and Ballinger, C. (2011). Exercise for improving balance in older people. Cochrane Database Syst Rev CD004963.

Hsiao-Wecksler, E.T. (2008). Biomechanical and age-related differences in balance recovery using the tether-release method. J Electromyogr Kinesiol *18*, 179–187.

Hsiao-Wecksler, E.T., and Robinovitch, S.N. (2007). The effect of step length on young and elderly women's ability to recover balance. Clin Biomech *22*, 574–580.

Hunter, G.R., McCarthy, J.P., and Bamman, M.M. (2004). Effects of resistance training on older adults. Sports Med. *34*, 329–348.

Hurley, M.V., Rees, J., and Newham, D.J. (1998). Quadriceps function, proprioceptive acuity and functional performance in healthy young, middle-aged and elderly subjects. Age Ageing *27*, 55–62.

Imamizu, H., Miyauchi, S., Tamada, T., Sasaki, Y., Takino, R., Pütz, B., Yoshioka, T., and Kawato, M. (2000). Human cerebellar activity reflecting an acquired internal model of a new tool. Nature *403*, 192–195.

Iqbal, K., and Pai, Y. (2000). Predicted region of stability for balance recovery: motion at the knee joint can improve termination of forward movement. J Biomech *33*, 1619–1627.

Islam, M.M., Nasu, E., Rogers, M.E., Koizumi, D., Rogers, N.L., and Takeshima, N. (2004). Effects of combined sensory and muscular training on balance in Japanese older adults. Prev Med *39*, 1148–1155.

Ivers, R.Q., Cumming, R.G., Mitchell, P., and Attebo, K. (1998). Visual impairment and falls in older adults: The Blue Mountains eye study. J Am Geriatr Soc *46*, 58–64.

Ivers, R.Q., Norton, R., Cumming, R.G., Butler, M., and Campbell, A.J. (2000). Visual impairment and risk of hip fracture. Am J Epidemiol *152*, 633–639.

Izquierdo, M., Ibañez, J., Gorostiaga, E., Garrues, M., Zúñiga, A., Antón, A., Larrión, J.L., and Häkkinen, K. (1999). Maximal strength and power characteristics in isometric and dynamic actions of the upper and lower extremities in middle-aged and older men. Acta Physiol. Scand. *167*, 57–68.

Jacobs, J.V., and Horak, F.B. (2007). Cortical control of postural responses. J Neural Transm *114*, 1339–1348.

Jensen, J.L., Brown, L.A., and Woollacott, M.H. (2001). Compensatory stepping: the biomechanics of a preferred response among older adults. Exp Aging Res *27*, 361–376.

Jöbges, M., Heuschkel, G., Pretzel, C., Illhardt, C., Renner, C., and Hummelsheim, H. (2004). Repetitive training of compensatory steps: a therapeutic approach for postural instability in Parkinson's disease. J Neurol Neurosurg Psychiatr *75*, 1682–1687.

Judge, J.O., King, M.B., Whipple, R., Clive, J., and Wolfson, L.I. (1995). Dynamic balance in older persons: effects of reduced visual and proprioceptive input. J Gerontol A Biol Sci Med Sci *50*, M263–270.

Juraska, J.M., and Lowry, N.C. (2012). Neuroanatomical changes associated with cognitive aging. Curr Top Behav Neurosci *10*, 137–162.

Kaasinen, V., and Rinne, J.O. (2002). Functional imaging studies of dopamine system and cognition in normal aging and Parkinson's disease. Neurosci Biobehav Rev *26*, 785–793.

Kamen, G., Sison, S.V., Du, C.C., and Patten, C. (1995). Motor unit discharge behavior in older adults during maximal-effort contractions. J Appl Physiol *79*, 1908–1913.

Kang, H.G., and Dingwell, J.B. (2008). Separating the effects of age and walking speed on gait variability. Gait Posture *27*, 572–577.

Kang, H.G., and Dingwell, J.B. (2009). Dynamics and stability of muscle activations during walking in healthy young and older adults. J Biomech *42*, 2231–2237.

Karamanidis, K., and Arampatzis, A. (2005). Mechanical and morphological properties of different muscle-tendon units in the lower extremity and running mechanics: effect of aging and physical activity. J Exp Biol *208*, 3907–3923.

Karamanidis, K., and Arampatzis, A. (2007). Age-related degeneration in leg-extensor muscle-tendon units decreases recovery performance after a forward fall: compensation with running experience. Eur J Appl Physiol *99*, 73–85.

Karamanidis, K., Arampatzis, A., and Mademli, L. (2008). Age-related deficit in dynamic stability control after forward falls is affected by muscle strength and tendon stiffness. J Electromyogr Kinesiol *18*, 980–989.

Kelly, V.E., Janke, A.A., and Shumway-Cook, A. (2010). Effects of instructed focus and task difficulty on concurrent walking and cognitive task performance in healthy young adults. Exp Brain Res *207*, 65–73.

Kerrigan, D.C., Todd, M.K., Della Croce, U., Lipsitz, L.A., and Collins, J.J. (1998). Biomechanical gait alterations independent of speed in the healthy elderly: evidence for specific limiting impairments. Arch Phys Med Rehabil *79*, 317–322.

Keshner, E.A., and Cohen, H. (1989). Current concepts of the vestibular system reviewed: 1. The role of the vestibulospinal system in postural control. Am J Occup Ther *43*, 320–330.

Keysor, J.J., and Jette, A.M. (2001). Have we oversold the benefit of late-life exercise? J Gerontol A Biol Sci Med Sci *56*, M412–423.

King, M.B., and Tinetti, M.E. (1995). Falls in community-dwelling older persons. J Am Geriatr Soc *43*, 1146–1154.

Klass, M., Baudry, S., and Duchateau, J. (2008). Age-related decline in rate of torque development is accompanied by lower maximal motor unit discharge frequency during fast contractions. J Appl Physiol *104*, 739–746.

Koceja, D.M., and Mynark, R.G. (2000). Comparison of heteronymous monosynaptic Ia facilitation in young and elderly subjects in supine and standing positions. Int J Neurosci *103*, 1–17.

Kovacs, C.R. (2005). Age-Related Changes in Gait and Obstacle Avoidance Capabilities in Older Adults: A Review. J Appl Gerontol *24*, 21 –34.

Lajoie, Y., and Gallagher, S.P. (2004). Predicting falls within the elderly community: comparison of postural sway, reaction time, the Berg balance scale and the Activities-specific Balance Confidence (ABC) scale for comparing fallers and non-fallers. Arch Gerontol Geriatr *38*, 11–26.

Lajoie, Y., Girard, A., and Guay, M. (2002). Comparison of the reaction time, the Berg Scale and the ABC in non-fallers and fallers. Arch Gerontol Geriatr *35*, 215–225.

Lajoie, Y., Teasdale, N., Bard, C., and Fleury, M. (1996). Upright standing and gait: are there changes in attentional requirements related to normal aging? Exp Aging Res *22*, 185–198.

Landi, F., Liperoti, R., Russo, A., Giovannini, S., Tosato, M., Capoluongo, E., Bernabei, R., and Onder, G. (2012). Sarcopenia as a risk factor for falls in elderly individuals: Results from the ilSIRENTE study. Clin Nutr (Edinburgh, Scotland) *31*, 652-8.

Larish, D.D., Martin, P.E., and Mungiole, M. (1988). Characteristic patterns of gait in the healthy old. Ann N Y Acad Sci *515*, 18–32.

Larsson, L., Grimby, G., and Karlsson, J. (1979). Muscle strength and speed of movement in relation to age and muscle morphology. J Appl Physiol *46*, 451 –456.

Latham, N.K., Bennett, D.A., Stretton, C.M., and Anderson, C.S. (2004). Systematic review of progressive resistance strength training in older adults. J Gerontol A Biol Sci Med Sci *59*, 48–61.

Lexell, J. (1995). Human Aging, Muscle Mass, and Fiber Type Composition. J Gerontol A Biol Sci Med Sci *50A*, 11–16.

Lexell, J., Taylor, C.C., and Sjöström, M. (1988). What is the cause of the ageing atrophy? Total number, size and proportion of different fiber types studied in whole vastus lateralis muscle from 15- to 83-year-old men. J Neurol Sci *84*, 275–294.

Li, K.Z., Lindenberger, U., Freund, A.M., and Baltes, P.B. (2001). Walking while memorizing: age-related differences in compensatory behavior. Psychol Sci *12*, 230–237.

Light, K.E. (1990). Information processing for motor performance in aging adults. Phys Ther *70*, 820–826.

van der Linden, M.H., Hendricks, H.T., Bloem, B.R., and Duysens, J. (2009). Hitting a support surface at unexpected height during walking induces loading transients. Gait Posture *29*, 255–260.

Lindenberger, U., Marsiske, M., and Baltes, P.B. (2000). Memorizing while walking: increase in dual-task costs from young adulthood to old age. Psychol Aging *15*, 417–436.

Llewellyn, M., Yang, J.F., and Prochazka, A. (1990). Human H-reflexes are smaller in difficult beam walking than in normal treadmill walking. Exp Brain Res *83*, 22–28.

Lockhart, T.E. (2008). An integrated approach towards identifying age-related mechanisms of slip initiated falls. J Electromyogr Kinesiol *18*, 205–217.

Lockhart, T.E., Smith, J.L., and Woldstad, J.C. (2005). Effects of aging on the biomechanics of slips and falls. Hum Factors *47*, 708–729.

Lockhart, T.E., Spaulding, J.M., and Park, S.H. (2007). Age-related slip avoidance strategy while walking over a known slippery floor surface. Gait Posture *26*, 142–149.

Lockhart, T.E., Woldstad, J.C., and Smith, J.L. (2003). Effects of age-related gait changes on the biomechanics of slips and falls. Ergonomics *46*, 1136–1160.

Lord, S.R., Clark, R.D., and Webster, I.W. (1991). Postural stability and associated physiological factors in a population of aged persons. J Gerontol *46*, M69–76.

Lord, S.R., and Dayhew, J. (2001). Visual Risk Factors for Falls in Older People. J Am Geriatr Soc *49*, 508.

Lord, S.R., Lloyd, D.G., and Li, S.K. (1996). Sensori-motor function, gait patterns and falls in community-dwelling women. Age Ageing *25*, 292–299.

Lord, S.R., McLean, D., and Stathers, G. (1992). Physiological factors associated with injurious falls in older people living in the community. Gerontology *38*, 338–346.

Lord, S.R., Rogers, M.W., Howland, A., and Fitzpatrick, R. (1999). Lateral stability, sensorimotor function and falls in older people. J Am Geriatr Soc *47*, 1077–1081.

Lord, S.R., Ward, J.A., Williams, P., and Anstey, K.J. (1993). An epidemiological study of falls in older community-dwelling women: the Randwick falls and fractures study. Aust J Public Health *17*, 240–245.

Luchies, C.W., Wallace, D., Pazdur, R., Young, S., and DeYoung, A.J. (1999). Effects of age on balance assessment using voluntary and involuntary step tasks. J Gerontol A Biol Sci Med Sci *54*, M140–144.

Lythgo, N., Begg, R., and Best, R. (2007). Stepping responses made by elderly and young female adults to approach and accommodate known surface height changes. Gait Posture *26*, 82–89.

Macaluso, A., Nimmo, M.A., Foster, J.E., Cockburn, M., McMillan, N.C., and De Vito, G. (2002). Contractile muscle volume and agonist-antagonist coactivation account for differences in torque between young and older women. Muscle Nerve *25*, 858–863.

Mackey, D.C., and Robinovitch, S.N. (2005). Postural steadiness during quiet stance does not associate with ability to recover balance in older women. Clin Biomech (Bristol, Avon) *20*, 776–783.

MacLellan, M.J., and Patla, A.E. (2006). Adaptations of walking pattern on a compliant surface to regulate dynamic stability. Exp Brain Res *173*, 521–530.

Mademli, L., Arampatzis, A., and Karamanidis, K. (2008). Dynamic stability control in forward falls: postural corrections after muscle fatigue in young and older adults. Eur J Appl Physiol. *103*, 295–306.

Madigan, M.L., and Lloyd, E.M. (2005). Age and stepping limb performance differences during a single-step recovery from a forward fall. J Gerontol A Biol Sci Med Sci *60*, 481–485.

Madureira, M.M., Takayama, L., Gallinaro, A.L., Caparbo, V.F., Costa, R.A., and Pereira, R.M.R. (2007). Balance training program is highly effective in improving functional status and reducing the risk of falls in elderly women with osteoporosis: a randomized controlled trial. Osteoporos Int *18*, 419–425.

Magill, R.A. (2011). Motor Learning and Control. Concepts and Applications (9th International Edition 2011; New York: McGraw-Hill).

Maki, B.E. (1997). Gait changes in older adults: predictors of falls or indicators of fear. J Am Geriatr Soc *45*, 313–320.

Maki, B.E., Edmondstone, M.A., and McIlroy, W.E. (2000). Age-related differences in laterally directed compensatory stepping behavior. J Gerontol A Biol Sci Med Sci *55*, M270–277.

Maki, B.E., and McIlroy, W.E. (1996). Influence of arousal and attention on the control of postural sway. J Vestib Res *6*, 53–59.

Maki, B.E., and McIlroy, W.E. (1997). The role of limb movements in maintaining upright stance: the "change-in-support" strategy. Phys Ther *77*, 488–507.

Maki, B.E., and McIlroy, W.E. (1999). The control of foot placement during compensatory stepping reactions: does speed of response take precedence over stability? IEEE Trans Rehabil Eng *7*, 80–90.

Maki, B.E., and McIlroy, W.E. (2006). Control of rapid limb movements for balance recovery: age-related changes and implications for fall prevention. Age Ageing *35 Suppl 2*, ii12–ii18.

Maki, B.E., Zecevic, A., Bateni, H., Kirshenbaum, N., and McIlroy, W.E. (2001). Cognitive demands of executing postural reactions: does aging impede attention switching? Neuroreport *12*, 3583–3587.

Malone, L.A., and Bastian, A.J. (2010). Thinking about walking: effects of conscious correction versus distraction on locomotor adaptation. J Neurophysiol *103*, 1954–1962.

Manini, T.M., and Clark, B.C. (2011). Dynapenia and Aging: An Update. J Gerontol A Biol Sci Med Sci *67A*, 28–40.

Mansfield, A., Peters, A.L., Liu, B.A., and Maki, B.E. (2010). Effect of a perturbation-based balance training program on compensatory stepping and grasping reactions in older adults: a randomized controlled trial. Phys Ther *90*, 476–491.

Marigold, D.S., Eng, J.J., Dawson, A.S., Inglis, J.T., Harris, J.E., and Gylfadóttir, S. (2005). Exercise leads to faster postural reflexes, improved balance and mobility, and fewer falls in older persons with chronic stroke. J Am Geriatr Soc *53*, 416–423.

Marigold, D.S., and Patla, A.E. (2002). Strategies for dynamic stability during locomotion on a slippery surface: effects of prior experience and knowledge. J Neurophysiol *88*, 339–353.

Marigold, D.S., and Patla, A.E. (2008). Age-related changes in gait for multi-surface terrain. Gait Posture *27*, 689–696.

Marquis, S., Moore, M.M., Howieson, D.B., Sexton, G., Payami, H., Kaye, J.A., and Camicioli, R. (2002). Independent predictors of cognitive decline in healthy elderly persons. Arch Neurol *59*, 601–606.

Martin, T.A., Keating, J.G., Goodkin, H.P., Bastian, A.J., and Thach, W.T. (1996a). Throwing while looking through prisms. I. Focal olivocerebellar lesions impair adaptation. Brain *119 (Pt 4)*, 1183–1198.

Martin, T.A., Keating, J.G., Goodkin, H.P., Bastian, A.J., and Thach, W.T. (1996b). Throwing while looking through prisms. II. Specificity and storage of multiple gaze-throw calibrations. Brain *119 (Pt 4)*, 1199–1211.

Mattay, V.S., Fera, F., Tessitore, A., Hariri, A.R., Das, S., Callicott, J.H., and Weinberger, D.R. (2002). Neurophysiological correlates of age-related changes in human motor function. Neurology *58*, 630–635.

McAndrew, P.M., Wilken, J.M., and Dingwell, J.B. (2011). Dynamic stability of human walking in visually and mechanically destabilizing environments. J Biomech *44*, 644–649.

McChesney, J.W., and Woollacott, M.H. (2000). The effect of age-related declines in proprioception and total knee replacement on postural control. J Gerontol A Biol Sci Med Sci *55*, M658–666.

McGibbon, C.A. (2003). Toward a better understanding of gait changes with age and disablement: neuromuscular adaptation. Exerc Sport Sci Rev *31*, 102–108.

McIlroy, W.E., and Maki, B.E. (1996). Age-related changes in compensatory stepping in response to unpredictable perturbations. J Gerontol A Biol Sci Med Sci *51*, M289–296.

McIlroy, W.E., Norrie, R.G., Brooke, J.D., Bishop, D.C., Nelson, A.J., and Maki, B.E. (1999). Temporal properties of attention sharing consequent to disturbed balance. Neuroreport *10*, 2895–2899.

Mcllroy, W., and Maki, B. (1995). Adaptive changes to compensatory stepping responses. Gait Posture *3*, 43–50.

McNay, E.C., and Willingham, D.B. (1998). Deficit in learning of a motor skill requiring strategy, but not of perceptuomotor recalibration, with aging. Learn Mem *4*, 411–420.

Menz, H.B., Lord, S.R., and Fitzpatrick, R.C. (2003). Age-related differences in walking stability. Age Ageing *32*, 137–142.

Minetti, A.E., Capelli, C., Zamparo, P., di Prampero, P.E., and Saibene, F. (1995). Effects of stride frequency on mechanical power and energy expenditure of walking. Med Sci Sports Exerc *27*, 1194–1202.

Mirelman, A., Herman, T., Brozgol, M., Dorfman, M., Sprecher, E., Schweiger, A., Giladi, N., and Hausdorff, J.M. (2012). Executive function and falls in older adults: new findings from a five-year prospective study link fall risk to cognition. PLoS ONE 7, e40297.

Miwa, T., Miwa, Y., and Kanda, K. (1995). Dynamic and static sensitivities of muscle spindle primary endings in aged rats to ramp stretch. Neurosci Lett *201*, 179–182.

Moreland, J.D., Richardson, J.A., Goldsmith, C.H., and Clase, C.M. (2004). Muscle weakness and falls in older adults: a systematic review and meta-analysis. J Am Geriatr Soc *52*, 1121–1129.

Morita, H., Shindo, M., Yanagawa, S., Yoshida, T., Momoi, H., and Yanagisawa, N. (1995). Progressive decrease in heteronymous monosynaptic Ia facilitation with human ageing. Exp Brain Res *104*, 167–170.

Morton, S.M., and Bastian, A.J. (2006). Cerebellar contributions to locomotor adaptations during splitbelt treadmill walking. J Neurosci *26*, 9107–9116.

Morton, S.M., Lang, C.E., and Bastian, A.J. (2001). Inter- and intra-limb generalization of adaptation during catching. Exp Brain Res *141*, 438–445.

Mynark, R.G., and Koceja, D.M. (2001). Effects of Age on the Spinal Stretch Reflex. J Appl Biomech *17*, 188–203.

Mynark, R.G., and Koceja, D.M. (2002). Down training of the elderly soleus H reflex with the use of a spinally induced balance perturbation. J Appl Physiol *93*, 127–133.

Narici, M.V., Maganaris, C.N., Reeves, N.D., and Capodaglio, P. (2003). Effect of aging on human muscle architecture. J Appl Physiol *95*, 2229–2234.

Nashner, L.M. (1980). Balance adjustments of humans perturbed while walking. J Neurophysiol *44*, 650–664.

Nashner, L.M. (1990). Sensory, neuromuscular and biomechanical contributions to human balance. In Duncan, P.W. (Ed.): Balance: Proceedings of the APTA Forum, American Physical Therapy Association, (Alexandria, VA: American Physical Therapy Association).

Neptune, R.R., and Sasaki, K. (2005). Ankle plantar flexor force production is an important determinant of the preferred walk-to-run transition speed. J Exp Biol *208*, 799–808.

Nitz, J.C., and Choy, N.L. (2004). The efficacy of a specific balance-strategy training programme for preventing falls among older people: a pilot randomised controlled trial. Age Ageing *33*, 52–58.

Norrie, R.G., Maki, B.E., Staines, W.R., and McIlroy, W.E. (2002). The time course of attention shifts following perturbation of upright stance. Exp Brain Res *146*, 315–321.

Onambele, G.L., Narici, M.V., and Maganaris, C.N. (2006). Calf muscle-tendon properties and postural balance in old age. J Appl Physiol *100*, 2048–2056.

Van Ooteghem, K., Frank, J.S., and Horak, F.B. (2009). Practice-related improvements in posture control differ between young and older adults exposed to continuous, variable amplitude oscillations of the support surface. Exp Brain Res *199*, 185–193.

Owings, T.M., Pavol, M.J., Foley, K.T., and Grabiner, M.D. (2000). Measures of postural stability are not predictors of recovery from large postural disturbances in healthy older adults. J Am Geriatr Soc *48*, 42–50.

Pai, Y.-C. (2003). Movement termination and stability in standing. Exerc Sport Sci Rev *31*, 19–25.

Pai, Y.-C., Bhatt, T., Wang, E., Espy, D., and Pavol, M.J. (2010). Inoculation against falls: rapid adaptation by young and older adults to slips during daily activities. Arch Phys Med Rehabil *91*, 452–459.

Pai, Y.C., and Iqbal, K. (1999). Simulated movement termination for balance recovery: can movement strategies be sought to maintain stability in the presence of slipping or forced sliding? J Biomech *32*, 779–786.

Pai, Y.C., and Patton, J. (1997). Center of mass velocity-position predictions for balance control. J Biomech *30*, 347–354.

Pai, Y.-C., Wening, J.D., Runtz, E.F., Iqbal, K., and Pavol, M.J. (2003). Role of feedforward control of movement stability in reducing slip-related balance loss and falls among older adults. J Neurophysiol *90*, 755–762.

Park, D.C., and Reuter-Lorenz, P. (2009). The adaptive brain: aging and neurocognitive scaffolding. Annu Rev Psychol *60*, 173–196.

Patel, M., Magnusson, M., Kristinsdottir, E., and Fransson, P.-A. (2009). The contribution of mechanoreceptive sensation on stability and adaptation in the young and elderly. Eur J Appl Physiol *105*, 167–173.

Patla, A., Frank, J., and Winter, D. (1990). Assessment of balance control in the elderly: major issues. *42*, 89–97.

Patla AE. Age-related changes in visually guided locomotion over different terrains: major issues. In: Stelmach GE and Homberg V (eds.): Sensorimotor Impairment in the Elderly (Amsterdam, the Netherlands: Kluwer Academic Publishers; 231-252).

Patla, A.E. (2003). Strategies for dynamic stability during adaptive human locomotion. IEEE Eng Med Biol Mag *22*, 48–52.

Patla, A.E., Prentice, S.D., Robinson, C., and Neufeld, J. (1991). Visual control of locomotion: strategies for changing direction and for going over obstacles. J Exp Psychol Hum Percept Perform *17*, 603–634.

Pavol, M.J., Owings, T.M., Foley, K.T., and Grabiner, M.D. (2001). Mechanisms Leading to a Fall From an Induced Trip in Healthy Older Adults. J Gerontol A Biol Sci Med Sci *56*, M428 – M437.

Pavol, M.J., Runtz, E.F., Edwards, B.J., and Pai, Y.-C. (2002). Age influences the outcome of a slipping perturbation during initial but not repeated exposures. J Gerontol A Biol Sci Med Sci *57*, M496–503.

Pavol, M.J., Runtz, E.F., and Pai, Y.-C. (2004). Young and older adults exhibit proactive and reactive adaptations to repeated slip exposure. J Gerontol A Biol Sci Med Sci *59*, 494–502.

Perry, S.D. (2006). Evaluation of age-related plantar-surface insensitivity and onset age of advanced insensitivity in older adults using vibratory and touch sensation tests. Neurosci Lett *392*, 62–67.

Peterka, R.J. (2002). Sensorimotor integration in human postural control. J Neurophysiol *88*, 1097–1118.

Petrella, R.J., Lattanzio, P.J., and Nelson, M.G. (1997). Effect of age and activity on knee joint proprioception. Am J Phys Med Rehabil *76*, 235–241.

Pijnappels, M., Bobbert, M.F., and van Dieën, J.H. (2004). Contribution of the support limb in control of angular momentum after tripping. J Biomech *37*, 1811–1818.

Pijnappels, M., Bobbert, M.F., and Dieën, J.H. van (2005). Push-off reactions in recovery after tripping discriminate young subjects, older non-fallers and older fallers. Gait Posture *21*, 388–394.

Pijnappels, M., van der Burg, J.C.E., Reeves, N.D., and van Dieën, J.H. (2006). Identification of high-risk fallers by force capacity measures in the elderly. J Biomech *39, Suppl 1*, S87.

Prince, F., Corriveau, H., Hébert, R., and Winter, D.A. (1997). Gait in the elderly. Gait Posture *5*, 128–135.

Puttemans, V., Wenderoth, N., and Swinnen, S.P. (2005). Changes in brain activation during the acquisition of a multifrequency bimanual coordination task: from the cognitive stage to advanced levels of automaticity. J Neurosci *25*, 4270–4278.

Quintern, J., Berger, W., and Dietz, V. (1985). Compensatory reactions to gait perturbations in man: short- and long-term effects of neuronal adaptation. Neurosci Lett *62*, 371–376.

Raj, I.S., Bird, S.R., and Shield, A.J. (2010). Aging and the force-velocity relationship of muscles. Exp Gerontol *45*, 81–90.

Ramnani, N. (2006). The primate cortico-cerebellar system: anatomy and function. Nat Rev Neurosci *7*, 511–522.

Rankin, J.K., Woollacott, M.H., Shumway-Cook, A., and Brown, L.A. (2000). Cognitive influence on postural stability: a neuromuscular analysis in young and older adults. J Gerontol A Biol Sci Med Sci *55*, M112–119.

Raz, N., Gunning, F.M., Head, D., Dupuis, J.H., McQuain, J., Briggs, S.D., Loken, W.J., Thornton, A.E., and Acker, J.D. (1997). Selective aging of the human cerebral cortex observed in vivo: differential vulnerability of the prefrontal gray matter. Cereb Cortex *7*, 268–282.

Raz, N., Lindenberger, U., Rodrigue, K.M., Kennedy, K.M., Head, D., Williamson, A., Dahle, C., Gerstorf, D., and Acker, J.D. (2005). Regional brain changes in aging healthy adults: general trends, individual differences and modifiers. Cereb Cortex *15*, 1676–1689.

Raz, N., Torres, I.J., Spencer, W.D., White, K., and Acker, J.D. (1992). Age-related regional differences in cerebellar vermis observed in vivo. Arch Neurol *49*, 412–416.

Redding, G.M., and Wallace, B. (1996). Adaptive spatial alignment and strategic perceptual-motor control. J Exp Psychol Hum Percept Perform *22*, 379–394.

Redding, G.M., and Wallace, B. (2002). Strategic calibration and spatial alignment: a model from prism adaptation. J Mot Behav *34*, 126–138.

Ribeiro, F., and Oliveira, J. (2010). Effect of physical exercise and age on knee joint position sense. Arch Gerontol Geriatr *51*, 64–67.

Richards, J., Mulavara, A., and Bloomberg, J. (2007). The interplay between strategic and adaptive control mechanisms in plastic recalibration of locomotor function. Exp Brain Res *178*, 326–338.

Riecker, A., Gröschel, K., Ackermann, H., Steinbrink, C., Witte, O., and Kastrup, A. (2006). Functional significance of age-related differences in motor activation patterns. Neuroimage *32*, 1345–1354.

Robinovitch, S.N., Heller, B., Lui, A., and Cortez, J. (2002). Effect of strength and speed of torque development on balance recovery with the ankle strategy. J Neurophysiol *88*, 613–620.

Rogers, M.W., Hedman, L.D., Johnson, M.E., Cain, T.D., and Hanke, T.A. (2001). Lateral stability during forward-induced stepping for dynamic balance recovery in young and older adults. J Gerontol A Biol Sci Med Sci *56*, M589–594.

Rogers, M.W., Johnson, M.E., Martinez, K.M., Mille, M.-L., and Hedman, L.D. (2003). Step training improves the speed of voluntary step initiation in aging. J Gerontol A Biol Sci Med Sci *58*, 46–51.

Rogers, M.W., and Mille, M.-L. (2003). Lateral stability and falls in older people. Exerc Sport Sci Rev *31*, 182–187.

Roller, C.A., Cohen, H.S., Kimball, K.T., and Bloomberg, J.J. (2002). Effects of normal aging on visuo-motor plasticity. Neurobiol. Aging *23*, 117–123.

Roos, P.E., McGuigan, M.P., and Trewartha, G. (2010). The role of strategy selection, limb force capacity and limb positioning in successful trip recovery. Clin Biomech (Bristol, Avon) *25*, 873–878.

Rose, D.J., and Clark, S. (2000). Can the control of bodily orientation be significantly improved in a group of older adults with a history of falls? J Am Geriatr Soc *48*, 275–282.

Rosenstein, M.T., Collins, J.J., and De Luca, C.J. (1993). A practical method for calculating largest Lyapunov exponents from small data sets. Physica D *65*, 117–134.

Rubenstein, L.Z. (2006). Falls in older people: epidemiology, risk factors and strategies for prevention. Age Ageing *35 Suppl 2*, ii37–ii41.

Rubenstein, L.Z., Josephson, K.R., Trueblood, P.R., Loy, S., Harker, J.O., Pietruszka, F.M., and Robbins, A.S. (2000). Effects of a group exercise program on strength, mobility, and falls among fall-prone elderly men. J Gerontol A Biol Sci Med Sci *55*, M317–321.

Sakai, M., Shiba, Y., Sato, H., and Takahira, N. (2008). Motor Adaptation during Slip-Perturbed Gait in Older Adults. J Phys Ther Sci *20*, 109–115.

Salthouse, T.A. (1994). The aging of working memory. Neuropsychology *8*, 535–543.

Salthouse, T.A., and Somberg, B.L. (1982). Time-accuracy relationships in young and old adults. J Gerontol A Biol Sci Med Sci *37*, 349–353.

Schillings, A.M., Mulder, T., and Duysens, J. (2005). Stumbling over obstacles in older adults compared to young adults. J Neurophysiol *94*, 1158–1168.

Schillings, A.M., van Wezel, B. m. h., Mulder, T., and Duysens, J. (2000). Muscular Responses and Movement Strategies During Stumbling Over Obstacles. J Neurophysiol *83*, 2093 –2102.

van Schooten, K.S., Sloot, L.H., Bruijn, S.M., Kingma, H., Meijer, O.G., Pijnappels, M., and van Dieën, J.H. (2011). Sensitivity of trunk variability and stability measures to balance impairments induced by galvanic vestibular stimulation during gait. Gait Posture *33*, 656–660.

Schubert, M., Beck, S., Taube, W., Amtage, F., Faist, M., and Gruber, M. (2008). Balance training and ballistic strength training are associated with task-specific corticospinal adaptations. Eur J Neurosci *27*, 2007–2018.

Schultz, A.B. (1995). Muscle function and mobility biomechanics in the elderly: an overview of some recent research. J Gerontol A Biol Sci Med Sci *50 Spec No*, 60–63.

Scott, S.H. (2004). Optimal feedback control and the neural basis of volitional motor control. Nat Rev Neurosci *5*, 532–546.

Seidler, R.D. (2006). Differential effects of age on sequence learning and sensorimotor adaptation. Brain Res Bull *70*, 337–346.

Seidler, R.D., Bernard, J.A., Burutolu, T.B., Fling, B.W., Gordon, M.T., Gwin, J.T., Kwak, Y., and Lipps, D.B. (2010). Motor control and aging: links to age-related brain structural, functional, and biochemical effects. Neurosci Biobehav Rev *34*, 721–733.

Shadmehr, R., and Mussa-Ivaldi, F.A. (1994). Adaptive representation of dynamics during learning of a motor task. J Neurosci *14*, 3208–3224.

Shaw, F.E. (2002). Falls in cognitive impairment and dementia. Clin Geriatr Med *18*, 159-173.

Sherrington, C., Tiedemann, A., Fairhall, N., Close, J.C.T., and Lord, S.R. (2011). Exercise to prevent falls in older adults: an updated meta-analysis and best practice recommendations. N S W Public Health Bull *22*, 78–83.

Sherrington, C., Whitney, J.C., Lord, S.R., Herbert, R.D., Cumming, R.G., and Close, J.C.T. (2008). Effective Exercise for the Prevention of Falls: A Systematic Review and Meta-Analysis. J Am Geriatr Soc *56*, 2234–2243.

Shimada, H., Obuchi, S., Furuna, T., and Suzuki, T. (2004). New intervention program for preventing falls among frail elderly people: the effects of perturbed walking exercise using a bilateral separated treadmill. Am J Phys Med Rehabil *83*, 493–499.

Shumway-Cook, A., Brauer, S., and Woollacott, M. (2000). Predicting the probability for falls in community-dwelling older adults using the Timed Up & Go Test. Phys Ther *80*, 896–903.

Shumway-Cook, A., Gruber, W., Baldwin, M., and Liao, S. (1997). The Effect of Multidimensional Exercises on Balance, Mobility, and Fall Risk in Community-Dwelling Older Adults. Phys Ther *77*, 46–57.

Shumway-Cook, A., and Woollacott, M. (1995). Motor Control: Theory and Practical Applications. (Baltimore: Md: Williams & Wilkins).

Shumway-Cook, A., and Woollacott, M. (2000). Attentional demands and postural control: the effect of sensory context. J Gerontol A Biol Sci Med Sci *55*, M10–16.

Shumway-Cook, A., and Woollacott, M.H. (2007). Motor Control - Translating Research into Clinical Practice (Philadelphia: Lippincott Williams & Wilkins).

Silsupadol, P., Siu, K.-C., Shumway-Cook, A., and Woollacott, M.H. (2006). Training of balance under single- and dual-task conditions in older adults with balance impairment. Phys Ther *86*, 269–281.

Skelton, D.A., Young, A., Greig, C.A., and Malbut, K.E. (1995). Effects of resistance training on strength, power, and selected functional abilities of women aged 75 and older. J Am Geriatr Soc *43*, 1081–1087.

Spirduso, W.W., Francis, K., and MacRae, P. (1995). Physical Dimensions of Aging (Champaign, IL: Human Kinetics Pub Inc).

Spirduso, W.W., Francis, K., and MacRae, P. (2005). Physical Dimensions of Aging (Champaign, IL: Human Kinetics Pub Inc).

Springer, S., Giladi, N., Peretz, C., Yogev, G., Simon, E.S., and Hausdorff, J.M. (2006). Dual-tasking effects on gait variability: the role of aging, falls, and executive function. Mov Disord *21*, 950–957.

Statistische Ämter des Bundes und der Länder (2011). Demografischer Wandel in Deutschland. Heft 1. Bevölkerungs- und Haushaltsentwicklung im Bund und in den Ländern. Wiesbaden: Statistisches Bundesamt.

Steadman, J., Donaldson, N., and Kalra, L. (2003). A randomized controlled trial of an enhanced balance training program to improve mobility and reduce falls in elderly patients. J Am Geriatr Soc *51*, 847–852.

Stelmach, G.E., and Worringham, C.J. (1985). Sensorimotor deficits related to postural stability. Implications for falling in the elderly. Clin Geriatr Med *1*, 679–694.

Sturnieks, D.L., Menant, J., Vanrenterghem, J., Delbaere, K., Fitzpatrick, R.C., and Lord, S.R. (2012). Sensorimotor and neuropsychological correlates of force perturbations that induce stepping in older adults. Gait Posture *36*, 356–360.

Sturnieks, D.L., St George, R., and Lord, S.R. (2008). Balance disorders in the elderly. Neurophysiol Clin *38*, 467–478.

Sundermier, L., Woollacott, M.H., Jensen, J.L., and Moore, S. (1996). Postural sensitivity to visual flow in aging adults with and without balance problems. J Gerontol A Biol Sci Med Sci *51*, M45–52.

Swash, M., and Fox, K.P. (1972). The effect of age on human skeletal muscle. Studies of the morphology and innervation of muscle spindles. J Neurol Sci *16*, 417–432.

Tang, P.F., and Woollacott, M.H. (1998). Inefficient postural responses to unexpected slips during walking in older adults. J Gerontol A Biol Sci Med Sci *53*, M471–480.

Tang, P.F., and Woollacott, M.H. (1999). Phase-dependent modulation of proximal and distal postural responses to slips in young and older adults. J Gerontol A Biol Sci Med Sci *54*, M89–102.

Taube, W. (2012). Neuronale Mechanismen der posturalen Kontrolle und der Einfluss von Gleichgewichtstraining. Journal für Neurologie, Neurochirurgie und Psychiatrie *4294967295*, 6–6.

Taube, W., Gruber, M., Beck, S., Faist, M., Gollhofer, A., and Schubert, M. (2007). Cortical and spinal adaptations induced by balance training: correlation between stance stability and corticospinal activation. Acta Physiol (Oxf) *189*, 347–358.

Taube, W., Gruber, M., and Gollhofer, A. (2008). Spinal and supraspinal adaptations associated with balance training and their functional relevance. Acta Physiol (Oxf) *193*, 101–116.

Taube, W., Schubert, M., Gruber, M., Beck, S., Faist, M., and Gollhofer, A. (2006). Direct corticospinal pathways contribute to neuromuscular control of perturbed stance. J Appl Physiol *101*, 420–429.

Teasdale, N., Bard, C., LaRue, J., and Fleury, M. (1993). On the cognitive penetrability of posture control. Exp Aging Res *19*, 1–13.

Teasdale, N., Stelmach, G.E., and Breunig, A. (1991). Postural sway characteristics of the elderly under normal and altered visual and support surface conditions. J Gerontol A Biol Sci Med Sci *46*, B238–244.

Terao, S., Sobue, G., Hashizume, Y., Li, M., Inagaki, T., and Mitsuma, T. (1996). Age-related changes in human spinal ventral horn cells with special reference to the loss of small neurons in the intermediate zone: a quantitative analysis. Acta Neuropathol. *92*, 109–114.

Terrier, P., and Dériaz, O. (2011). Kinematic variability, fractal dynamics and local dynamic stability of treadmill walking. J Neuroeng Rehabil *8*, 12.

Thelen, D.G., Muriuki, M., James, J., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (2000). Muscle activities used by young and old adults when stepping to regain balance during a forward fall. J Electromyogr Kinesiol *10*, 93–101.

Thelen, D.G., Schultz, A.B., Alexander, N.B., and Ashton-Miller, J.A. (1996). Effects of age on rapid ankle torque development. J Gerontol A Biol Sci Med Sci *51*, M226–232.

Thelen, D.G., Wojcik, L.A., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (1997). Age differences in using a rapid step to regain balance during a forward fall. J Gerontol A Biol Sci Med Sci *52*, M8–13.

Thies, S.B., Richardson, J.K., and Ashton-Miller, J.A. (2005). Effects of surface irregularity and lighting on step variability during gait: a study in healthy young and older women. Gait Posture *22*, 26–31.

Thom, J.M., Morse, C.I., Birch, K.M., and Narici, M.V. (2005). Triceps surae muscle power, volume, and quality in older versus younger healthy men. J Gerontol A Biol Sci Med Sci *60*, 1111–1117.

Thom, J.M., Morse, C.I., Birch, K.M., and Narici, M.V. (2007). Influence of muscle architecture on the torque and power-velocity characteristics of young and elderly men. Eur J Appl Physiol *100*, 613–619.

Thoroughman, K.A., and Shadmehr, R. (1999). Electromyographic correlates of learning an internal model of reaching movements. J Neurosci *19*, 8573–8588.

Tideiksaar, R. (2000). Stürze und Sturzprävention: Assessment - Prävention - Management (Bern: Verlag Hans Huber).

Tinetti, M.E., Speechley, M., and Ginter, S.F. (1988). Risk factors for falls among elderly persons living in the community. N Engl J Med *319*, 1701–1707.

Torres-Oviedo, G., and Bastian, A.J. (2010). Seeing is believing: effects of visual contextual cues on learning and transfer of locomotor adaptation. J Neurosci *30*, 17015–17022.

Torres-Oviedo, G., Vasudevan, E., Malone, L., and Bastian, A.J. (2011). Locomotor adaptation. Prog Brain Res *191*, 65–74.

Trappe, S., Gallagher, P., Harber, M., Carrithers, J., Fluckey, J., and Trappe, T. (2003). Single muscle fibre contractile properties in young and old men and women. J Physiol (Lond.) *552*, 47–58.

Tseng, S.-C., Stanhope, S.J., and Morton, S.M. (2009). Impaired reactive stepping adjustments in older adults. J Gerontol A Biol Sci Med Sci *64*, 807–815.

Vandervoort, A.A. (2002). Aging of the human neuromuscular system. Muscle Nerve *25*, 17–25.

Vandervoort, A.A., and McComas, A.J. (1986). Contractile changes in opposing muscles of the human ankle joint with aging. J Appl Physiol *61*, 361–367.

Verdú, E., Ceballos, D., Vilches, J.J., and Navarro, X. (2000). Influence of aging on peripheral nerve function and regeneration. J Peripher Nerv Syst *5*, 191–208.

Verghese, J., Lipton, R.B., Hall, C.B., Kuslansky, G., Katz, M.J., and Buschke, H. (2002). Abnormality of gait as a predictor of non-Alzheimer's dementia. N Engl J Med *347*, 1761–1768.

Verrillo, R.T., Bolanowski, S.J., and Gescheider, G.A. (2002). Effect of aging on the subjective magnitude of vibration. Somatosens Mot Res *19*, 238–244.

Wang, T.-Y., Bhatt, T., Yang, F., and Pai, Y.-C. (2011). Generalization of motor adaptation to repeated-slip perturbation across tasks. Neuroscience *180*, 85–95.

West, R.L. (1996). An application of prefrontal cortex function theory to cognitive aging. Psychol Bull *120*, 272–292.

Whitney, S.L., Hudak, M.T., and Marchetti, G.F. (2000). The dynamic gait index relates to self-reported fall history in individuals with vestibular dysfunction. J Vestib Res *10*, 99–105.

Wijlhuizen, G.J. (2008). Physical activity and falls in older persons. Development of the Balance Control Difficulty Homeostasis Model (Leiden: Universiteit Leiden, Dissertation).

Winter, D. (1991). The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological (Waterloo: University of Waterloo Press).

Winter, D. (1995). Human balance and posture control during standing and walking. Gait Posture *3*, 193–214.

Winter, D.A., Patla, A.E., and Frank, J.S. (1990). Assessment of balance control in humans. Med Prog Technol *16*, 31–51.

Wojcik, L.A., Thelen, D.G., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (1999). Age and gender differences in single-step recovery from a forward fall. J Gerontol A Biol Sci Med Sci *54*, M44–50.

Wojcik, L.A., Thelen, D.G., Schultz, A.B., Ashton-Miller, J.A., and Alexander, N.B. (2001). Age and gender differences in peak lower extremity joint torques and ranges of motion used during single-step balance recovery from a forward fall. J Biomech *34*, 67–73.

Wolfson, L., Whipple, R., Derby, C., Judge, J., King, M., Amerman, P., Schmidt, J., and Smyers, D. (1996). Balance and strength training in older adults: intervention gains and Tai Chi maintenance. J Am Geriatr Soc *44*, 498–506.

Wolfson, L.I., Whipple, R., Amerman, P., Kaplan, J., and Kleinberg, A. (1985). Gait and balance in the elderly. Two functional capacities that link sensory and motor ability to falls. Clin Geriatr Med *1*, 649–659.

Woltring, H.J. (1986). A Fortran package for generalized, cross-validatory spline smoothing and differentiation. Adv Eng Software (1978) *8*, 104–113.

Woo, J., Ho, S.C., Lau, J., Chan, S.G., and Yuen, Y.K. (1995). Age-associated gait changes in the elderly: pathological or physiological? Neuroepidemiology *14*, 65–71.

Woollacott, M., and Shumway-Cook, A. (2002). Attention and the control of posture and gait: a review of an emerging area of research. Gait Posture *16*, 1–14.

Woollacott, M.H. (1989). Aging, posture control, and movement preparation. In Woollacott, M.H. and Shumway-Cook, A.: Development of Posture and Gait Across the Lifespan. (Columbia, SC: University of South Carolina Press).

Woollacott, M.H. (1993). Age-related changes in posture and movement. J Gerontol A Biol Sci Med Sci *48 Spec No*, 56–60.

Woollacott, M.H., Shumway-Cook, A., and Nashner, L.M. (1986). Aging and posture control: changes in sensory organization and muscular coordination. Int J Aging Hum Dev *23*, 97–114.

Woollacott, M.H., and Tang, P.F. (1997). Balance control during walking in the older adult: research and its implications. Phys Ther *77*, 646–660.

Wrisley, D.M., and Kumar, N.A. (2010). Functional gait assessment: concurrent, discriminative, and predictive validity in community-dwelling older adults. Phys Ther *90*, 761–773.

Yaffe, K., Barnes, D., Nevitt, M., Lui, L.Y., and Covinsky, K. (2001). A prospective study of physical activity and cognitive decline in elderly women: women who walk. Arch Intern Med *161*, 1703–1708.

Yang, F., Bhatt, T., and Pai, Y.-C. (2009). Role of stability and limb support in recovery against a fall following a novel slip induced in different daily activities. J Biomech *42*, 1903–1908.

Yogev-Seligmann, G., Hausdorff, J.M., Giladi, N., 2008. The role of executive function and attention in gait. Mov Disord *23*, 329–342; quiz 472.

Young, W.R., and Hollands, M.A. (2012). Evidence for age-related decline in visuomotor function and reactive stepping adjustments. Gait Posture *6*, 477-81

Zatsiorsky, V.M., and Seluyanov, V. (1983). The mass and inertial characteristics of main segments of the human body. In Biomechanics VIII-B, (Champaign, IL: Human Kinetics Publishers), pp. 1152–1159.